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*Development of a Computational Finite Element Tool for the Stress Analysis of the
Femur and Tibia*

by

Anthony Garky Au



A thesis submitted to the Faculty of Graduate Studies and Research in partial
fulfillment of the requirements for the degree of *Master of Science*.

*Department of Mechanical Engineering
&
Department of Biomedical Engineering*

Edmonton, Alberta
Fall 2003

University of Alberta

Faculty of Graduate Studies and Research

The undersigned certify they have read, and recommend to the Faculty of Graduate Studies and Research for acceptance, a thesis entitled **Development of a Computational Finite Element Tool for the Stress Analysis of the Femur and Tibia** submitted by **Anthony Garky Au** in partial fulfillment of the requirements for the degree of **Master of Science**.



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Abstract

In America, nearly 10% of the 100,000 anterior cruciate ligament (ACL) reconstructions and the 250,000 total knee replacements (TKR) performed yearly require revision surgery. A better understanding of mechanical stresses in knee bones can reduce the incidence of revision surgeries. A finite element (FE) model was developed to study stresses in intact and surgically altered femora and tibiae. Model geometry was based on commercially available composite bones. Orthotropic, heterogeneous bone properties and non-uniform distributed loading were incorporated. The appropriateness of the FE model was verified with experimental data from literature.

The application of this tool to tunnel placement in ACL reconstruction predicted stress-shielding at the postero-lateral region of the tunnel wall. Prolonged stress-shielding may cause bone resorption leading to tunnel enlargement, compromising the reconstruction. Button fixation stresses on the femoral cortex were noticeable for low levels of loading. With repeated compression, these stress levels may cause bone microdamage resulting in fatigue failure. Stresses resulting from four tibia FE prosthesis model designs were compared. An elliptical post was found to relieve post base compression more effectively than a tapered post. Prosthesis pegs relieved surrounding bone stress, which may cause bone resorption leading to aseptic loosening of the prosthesis.

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Au, A. G., Otto, D. D., Raso, V. J., Amirfazli, A., 2003. Development of a finite element tool for stress analysis of femur and tibia incorporating anatomically realistic mechanical properties. Proceedings of the Second MIT Conference on Computational Fluid and Solid Mechanics, Cambridge, MA, 1617-1621.

Au, A. G., Raso, V. J., Liggins, A. B., Otto, D. D., Amirfazli, A., 2003. A three-dimensional finite element stress analysis for tunnel placement and buttons in anterior cruciate ligament reconstructions. Journal of Biomechanics, submitted.

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Chapter 1: Introduction

This chapter provides the necessary background to understand this thesis work; specifically, a better understanding of the biomechanics as applied to the human knee. The first part discusses the motivation of this work; followed by the statements of the specific objectives and the scope of this thesis. A general description of the anatomy of the human knee is also given.

1.1 Motivation

The anterior cruciate ligament (ACL) is one of the most commonly damaged ligaments in the knee. Epidemiologic studies estimate that 1 in 3000 people in the American general population sustain ACL injuries (Malinzak et al., 2001). This figure corresponds to an overall injury rate approaching 100,000 annually in the United States. The average cost for surgical repair of an ACL tear is approximately \$11,500 (US), translating to over \$1 billion (US) annual cost. As the average age of the North American population is rising, the incidence of total knee arthroplasties is also rising. In 1998, approximately 250,000 total knee arthroplasties were performed in the United States (AAOS, 2001) at a cost of approximately \$25,000 (US) per procedure. With the increasing number of both procedures, revision surgery has also become increasingly common. In 1995, 8% of ACL reconstruction patients had unsatisfactory results (Wolf and Lemak, 2002); in 1998, 9% of total knee arthroplasties required revision surgery (AAOS, 2001). Currently (in 2003), 8 to 10 ACL reconstructions are performed weekly in Edmonton (which has a population of approximately 1,000,000 people) (Cinats, 2003), and approximately 2,000 total knee replacements (TKR) are performed annually (Otto, 2003). Reducing the incidence of revision surgery is necessary to reduce health care costs as well as to improve patient health.

Accurate stress analyses of the distal femur and proximal tibia are important for predicting potential failures in total knee replacements and knee reconstructions. Stress distribution is believed to be the main stimulus in adaptive bone remodeling process and mechano-biological phenomena at the bone-implant interface (Kowalczyk, 2003). An

accurate determination of bone stresses in the knee bones can help reduce the incidence of knee prosthesis loosening in TKRs and tunnel enlargement in ACL reconstructions. While surface stresses can be determined by the use of strain gauges, a limitation of these experimental methods is they cannot determine internal stress and strain distributions. By using the finite element method, internal stress and strain information can be collected efficiently. A computational finite element tool can be helpful in predicting stress and strain distributions in surgical procedures such as knee reconstructions and knee replacements, thereby aiding to adopt procedures or designs that can lead to better surgical outcomes, hence reducing the incidence of revision surgery, as well as optimizing the design of knee prostheses.

1.2 Objectives of Thesis

This thesis describes the complete developmental process of a tool to predict stresses in the femur and tibia using the finite element method. Strategies in finite element (FE) modeling, specifically material property assignments and loading conditions, were investigated to produce an accurate and high quality FE model.

The specific objectives of this thesis were:

- Establish a new method of applying material property distributions to FE models using published experimental data.
- Establish a new method of applying more physiologically representative loading to FE models using published experimental data.
- Verify that the FE models are accurate by performing an internal numerical analysis.
- Verify the accuracy of the FE models by comparing numerical simulation data with experimentally determined data.
- Demonstrate the utility of this tool in clinical applications such as tunnel placement in anterior cruciate ligament reconstructions and total knee replacements.

1.3 Scope of Thesis

This work focuses on the use of the finite element method to analyze the internal stresses of the human knee through the use of a computational finite element program. In the current phase of investigation, modeling was restricted to the principal bones of the tibiofemoral joint (i.e. the femur and the tibia). Ligaments, muscles, menisci, and articular cartilage and their associated loading were not included in the current model, but will be included in future versions of the model. The analysis of stress and strains focused on the immediate region of the knee joint, specifically the distal femur and proximal tibia. All loading on the model was considered static, and the subsequent stresses and strains were investigated at the macrostructural level.

This thesis presents a *method* of creating a finite element model to investigate stresses in the tibiofemoral bones. As such, technical efficiencies (e.g. computational efficiencies, adaptive meshing) of the finite element method were not the main focus in the development of this tool. The method is used to provide a better representation of the stress states in the bones by incorporating more representative geometry, material properties, and loading conditions.

1.4 Background

To gain an understanding of the basic anatomy of the human knee, the bones of the knee and the associated ligaments will be discussed.

1.4.1 Planes and Sections

Any slice through a three-dimensional object can be described with reference to three sectional planes (Fig 1.1). The transverse plane lies at right angles to the long axis of the body, dividing it into superior and inferior sections. A cut in this plane is called a transverse section. The coronal (or frontal) plane extends from side to side, dividing the body into anterior and posterior sections. The sagittal plane extends from front to back,

dividing the body into left and right sections. A cut that passes along the midsagittal line is a midsagittal section; a cut parallel to the midsagittal line is a parasagittal section.

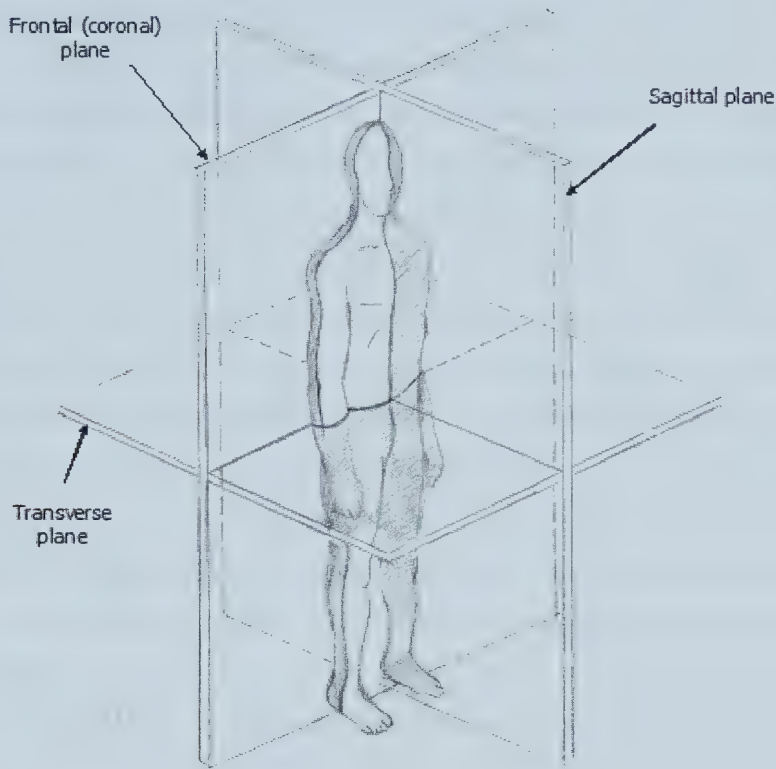


Figure 1.1: The three primary planes of section.

1.4.2 Directional Terms

Table 1.1 shows the directional terms that will be used extensively throughout this work.

Table 1.1: Anatomical Direction Terms

Term	Region
Anterior	The front
Posterior	The back
Medial	Toward the body's longitudinal axis
Lateral	Away from the body's longitudinal axis
Proximal	Toward an attached base
Distal	Away from an attached base
Superior	Above (toward the head)
Inferior	Below (away from the head)

1.4.3 Anatomy of the Knee

The knee is the largest and most complex joint in the human body. There are three major bones in the knee joint: the femur, the tibia, and the patella. The femur (Fig 1.2) is the longest and heaviest bone in the body and articulates with the tibia (Fig 1.3) of the leg at the knee joint. The shaft of the femur divides into the medial and lateral condyles as it approaches the knee joint. On the posterior surface, the two condyles are separated by a deep intercondylar fossa (or notch). On the anterior surface, the articular faces of the condyles merge, producing the patellar surface, over which the patella glides. The patella, also known as the kneecap, has a rough anterior surface which attaches to the quadriceps tendon as well as the patellar ligament. The articulating surfaces of the femur, patella, and tibia are covered by articular cartilage, which are slick and smooth to reduce friction during movement across the joint. When articular cartilage is “worn thin” due to osteoarthritis or rheumatoid arthritis, the contact between the femur and the tibia becomes painful. In cases of severe arthritic pain, a total knee replacement may be necessary to reduce the pain.

The medial and lateral condyles of the femur articulate with the medial and lateral tibial condyles at the proximal end of the tibia. The two condyles are separated by a ridge called the intercondylar eminence. The tibial tuberosity, marking the attachment of the patellar ligament, can easily be seen on the anterior surface of the tibia. The anterior crest is a ridge that begins at the tibial tuberosity and extends distally along the anterior surface.

There are four major ligaments connecting the femur and the tibia: the anterior cruciate ligament, the posterior cruciate ligament, the medial collateral ligament, and the lateral collateral ligament (Fig 1.4). The cruciate ligaments attach the intercondylar area of the tibia to the condyles of the femur, with anterior and posterior referring to the sites of origin. These ligaments limit the anterior and posterior movement of the femur and maintain the alignment of the femoral and tibial condyles. The primary role of the anterior cruciate ligament is to prevent anterior translation of the femur. The collateral ligaments reinforce the medial and lateral surfaces of the knee joint, and tighten only at full extension to stabilize the joint. The anterior cruciate ligament is relatively weak

compared to the posterior cruciate, medial and lateral collateral ligaments (Jarvela, 2001). The ACL has an ultimate tensile load of 2160 N (Woo et al., 1991), and is only half as strong as the posterior cruciate ligament (Kennedy et al., 1976). Damage of the anterior cruciate ligament is a commonly seen knee injury (Malinzak et al., 2001). Unlike the medial and lateral collateral ligaments, the anterior cruciate ligament has poor healing potential and must be treated upon injury (Fu et al., 1999). The most effective treatment of ACL injury is surgical reconstruction of the torn ligament.

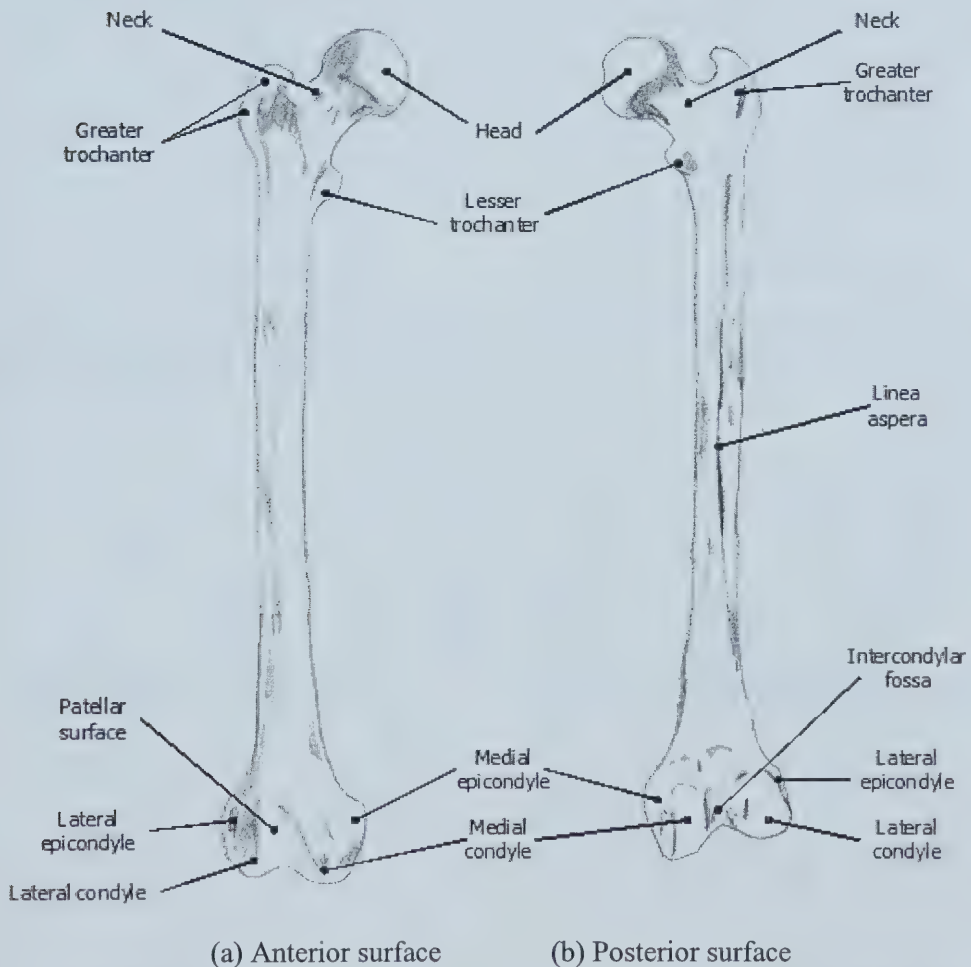


Figure 1.2: Anatomy of the human right femur with markings on (a) the anterior surface and (b) the posterior surface.

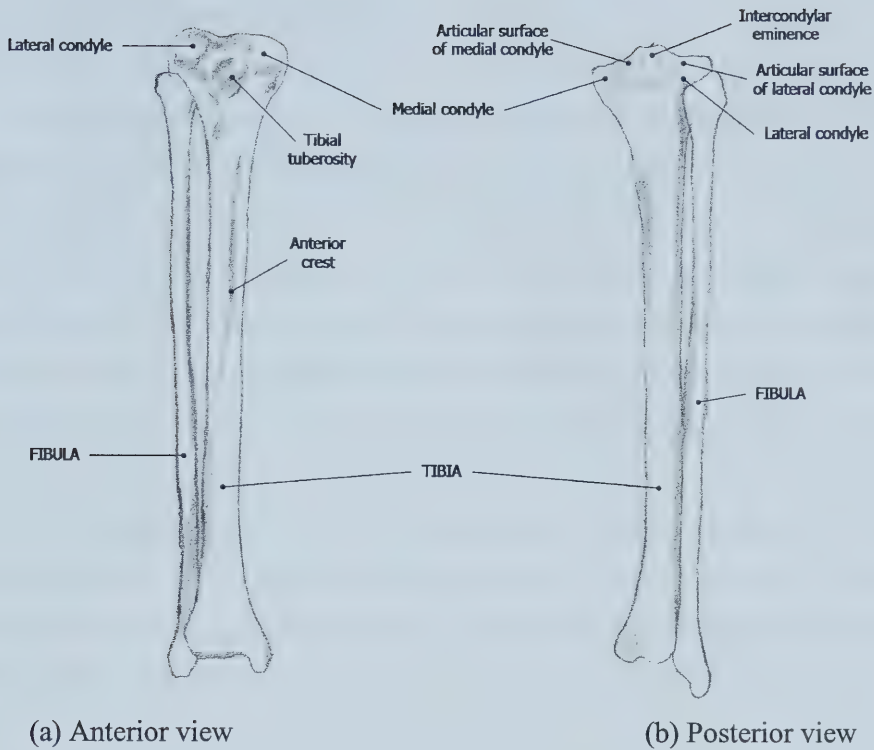


Figure 1.3: Anatomy of the human tibia. (a) Anterior view of the right tibia and fibula; (b) posterior view of the right tibia and fibula.

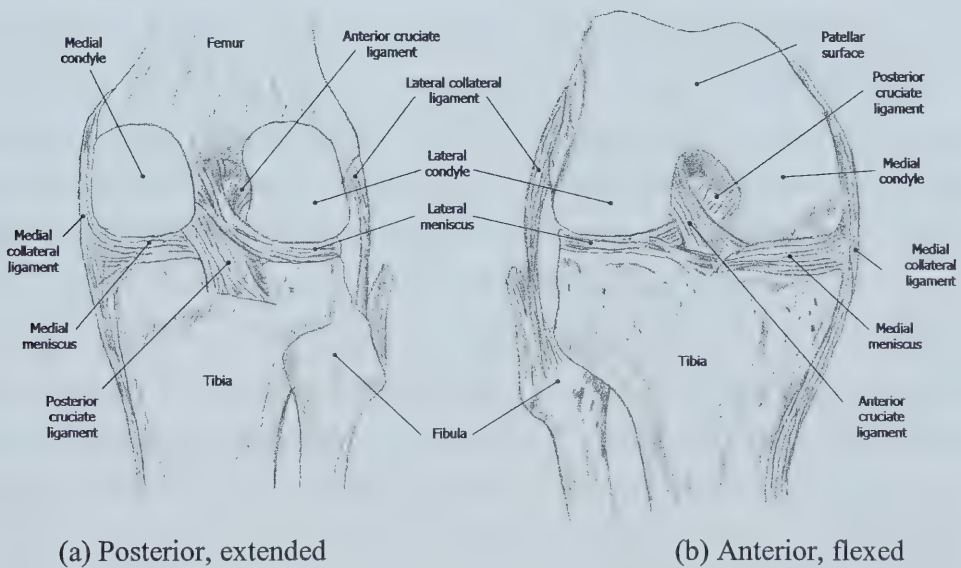


Figure 1.4: The ligaments of the knee joint. (a) Posterior view of the right knee at full extension. (b) Anterior view of the right knee at full flexion.

1.4.4 The Finite Element Method

An analytical solution is a mathematical expression that describes the unknown quantity at any location in a solution domain. Ideally, every problem should be solved using analytical methods. However, analytical methods often cannot be used to solve problems with irregular geometries and non-uniform material properties, among other classes of problems. Instead, numerical methods are more appropriate for solving such problems. A variety of numerical methods are available, each specialized for use in different classes of problems. The finite difference method is one numerical method that traditionally was used for heat transfer and fluid flow problems. The method is limited by the fact that regular boundaries (i.e. not arbitrary or curved) must exist. The boundary element method uses Green's theorem to transform a 3D volume integral to a 2D surface integral (or a 2D area to a 1D line integral). It is useful for small problems with infinite or semi-infinite domains (e.g. acoustic, underwater, soil mechanics, etc.) and homogeneous properties. It is sometimes better suited than the finite element method (FEM) because the FEM requires discretization of the entire domain. The finite element method is applicable to a wide range of boundary value problems in engineering. It has developed simultaneously with the increasing use of computers and the growing emphasis on numerical methods for engineering analysis. Specific engineering applications include structural engineering, soil mechanics, thermodynamics, fluid dynamics, and nuclear engineering. The FEM is particularly well-suited to the study of knee biomechanics because of its capability to evaluate stresses in structures of complex shape, loading, and material behaviour. It has gained considerable use in evaluating artificial joint replacements and fracture fixation (Huiskes and Chao, 1983).

The mathematical foundation for the finite element method (FEM) was laid down over a hundred years ago by mathematicians such as Rayleigh, Ritz, Galerkin, and Courant. The first paper outlining FEM, titled "Stiffness and Deflection Analysis of Complex Structures" was presented by Turner, Clough, Martin, and Topp in 1956. However, the phrase "finite element" was not coined until 1960 in a work by Clough. The finite element method gained momentum in the 1960s and, by 1990, over 50,000 articles have been published on the finite element method. The numerous books written

regarding the finite element method include those by Rao (1982), Norrie and De Vries (1973, 1978), and Desai and Abel (1972).

The basis of the finite element (FE) method is the representation of a body (or structure) by smaller subdivisions called finite elements. These elements are interconnected at nodal points, with simple functions approximating the actual displacement (in stress analysis problems) over each finite element. Using a variational principle of mechanics, a set of equilibrium equations is obtained for each element. The equilibrium equations for the entire body are obtained by combining the equations for the individual elements. These equations are modified for a desired set of boundary conditions and then solved for the unknown displacements. Additional manipulations can be performed to obtain stress and strain information. The finite element procedure is briefly elaborated upon in the following six sections.

Discretization of the continuum

The discretization of the solution region into finite elements is the first step in FEM. The step basically involves replacing a solution region having an infinite number of degrees of freedom with a system having a finite number of degrees of freedom. The finite elements may be triangles or quadrilaterals (for a 2D continuum) or tetrahedra or hexahedra (for a 3D continuum). The process of discretization is essentially at the judgement of the engineer, who must decide the appropriate number, size, and arrangement of finite elements that will give an effective representation of the given continuum for the particular problem considered.

Problems involving plane strain, plane stress, and plate bending can be approximated by a two-dimensional formulation. The basic elements used for 2D analysis are the triangle and the quadrilateral (or its specialized form, the rectangle). As the long bones of the lower limb are asymmetric in three planes, it is more representative to model them using 3D elements. The basic 3D elements are the tetrahedron (with 4 primary external nodes) and the hexahedron (with 8 primary external nodes) (Fig 1.5). These elements only have nodes on the corners and their edges are straight. However, due to the natural curvature of the bones, finite elements with curved sides are better

suited for modeling bones than straight-sided elements. This is made possible by the addition of secondary external nodes or midside nodes. Elements that have been considered for use in this study include the 4-node tetrahedral, 10-node tetrahedral, 8-node brick, and 20-node brick elements.

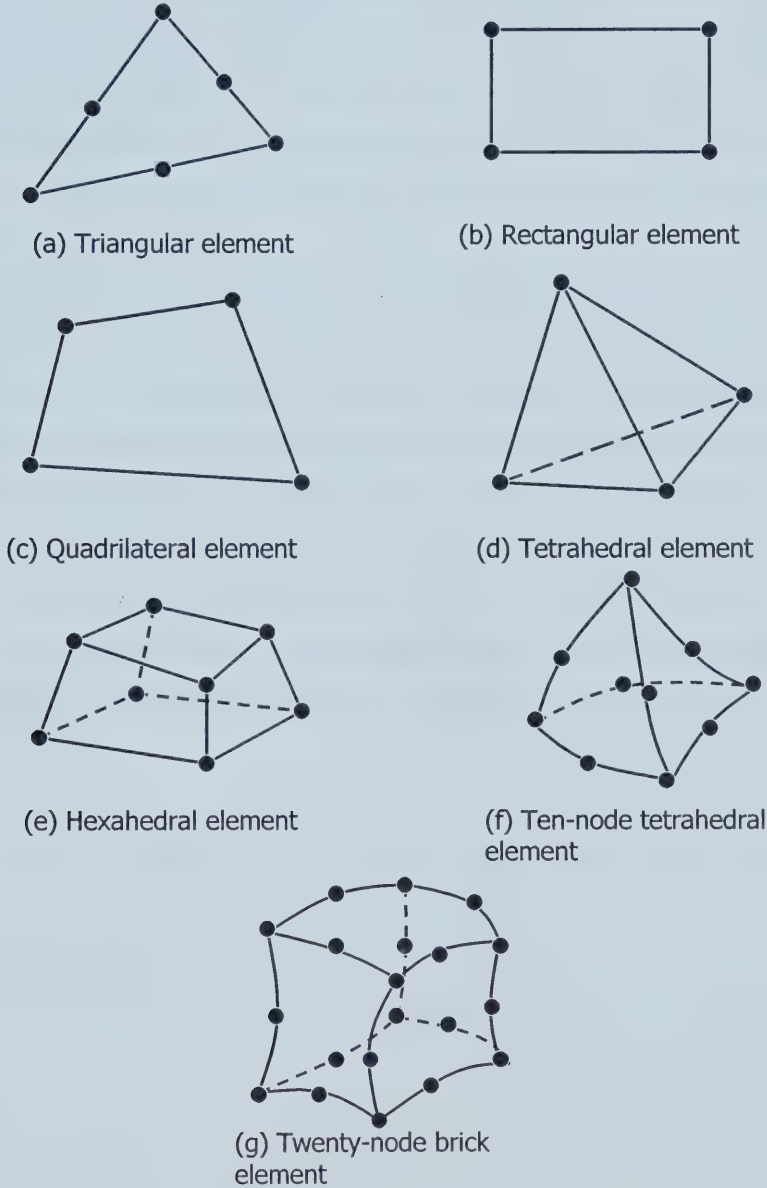


Figure 1.5: Two- and three-dimensional elements.

Selection of the displacement models

After careful selection of the element type to be used in the analysis, a displacement function must be selected. Displacement functions are used to approximate the distribution of the actual displacement over each finite element. Three interrelated factors influence the selection of a displacement model. The first is the degree of the displacement model. The second is the particular displacement magnitudes that describe the model. These are usually the displacements of the nodal points. The third factor is that the model should satisfy convergence requirements to ensure that numerical results approach the correct solution.

The displacement method of structural analysis involves dividing the structure into finite elements and using displacement functions to approximate the displacements for each element. A polynomial is the most common form of displacement model because (1) it is easier to perform differentiation or integration with polynomials, and (2) it is possible to improve the accuracy of the results by increasing the order of the polynomial. While a polynomial with infinite terms may be needed to represent the exact solution, practicality limits the number of terms that can be retained in the polynomial. The most common displacement models are the linear and quadratic polynomials. In this study, a quadratic 3D displacement model was selected. This polynomial is represented as:

$$u(x, y, z) = \alpha_1 + \alpha_2 x + \alpha_3 y + \alpha_4 z + \alpha_5 x^2 + \alpha_6 y^2 + \alpha_7 z^2 + \alpha_8 xy + \alpha_9 yz + \alpha_{10} xz \quad (1.1)$$

Or in matrix form:

$$u(x) = \{\phi\} \{\alpha\} \quad (1.2)$$

$$\{\phi\} = [1 \quad x \quad y \dots \quad xz]$$

$$\{\alpha\}^T = [\alpha_1 \quad \alpha_2 \quad \alpha_3 \dots \quad \alpha_{10}]$$

where u is the displacement in the three dimensions of x , y , and z ; and α_n are the generalized coordinates of the polynomial.

The basic idea of the FEM is to consider a body as composed of several elements connected at specified nodal points. The displacement inside any finite element is assumed to be given by a simple function in terms of nodal values of that element. The nodal values of the solution (also known as nodal degrees of freedom) are unknown. The solution of the system of equations (i.e. force equilibrium equations) gives the values for those unknown degrees of freedom. Once the nodal degrees of freedom are known, the solution within any finite element will also be known. Therefore, we need to express the approximating polynomial in terms of the nodal degrees of freedom of a typical element.

$$\{u\} = [\phi] \{\alpha\} \quad (1.3)$$

$$\{q\} = \begin{Bmatrix} u(\text{node } 1) \\ u(\text{node } 2) \\ \dots \\ u(\text{node } M) \end{Bmatrix} = \begin{bmatrix} u(\text{node } 1) \\ u(\text{node } 2) \\ \dots \\ u(\text{node } M) \end{bmatrix} \{\alpha\} = [A] \{\alpha\} \quad (1.4)$$

where M is the total number of nodes of the element, $\{q\}$ is the vector of nodal displacements, and $[A]$ is the transformation relating generalized and nodal displacements. To eliminate the generalized coordinates, we invert equation (1.4) to give

$$\{\alpha\} = [A^{-1}] \{q\} \quad (1.5)$$

where $[A^{-1}]$ is the displacement transformation matrix, and substitute equation (1.5) into equation (1.3), and obtain

$$\{u\} = [\phi][A^{-1}]\{q\} = [N]\{q\} \quad (1.6)$$

Equation (1.6) expresses the displacements $\{u\}$ at any point within the element in terms of the displacements at the nodes $\{q\}$. However, it is not always possible to obtain $[A^{-1}]$. The interpolation function representation of displacement models can avoid this difficulty, if the matrix $[N]$ can be directly constructed. Interpolation functions (or shape functions) are functions used to represent the behaviour of the solution within an element. They have a unit value at one nodal point, and zero values at all other nodal points. For each displacement in the vector $\{u\}$, we need an interpolation function corresponding to every nodal degree of freedom represented in the vector $\{q\}$.

In the finite element method, the numerical solution must converge or tend to the exact solution of the problem. In order for convergence to occur, three conditions must be met. First, the displacement models must be continuous within the elements, and the

displacements must be compatible between adjacent elements. Second, the displacement models must include the rigid body displacements of the element. This condition states that there exist combinations of values of the generalized coordinates that cause all points on the element to experience the same displacement. In the displacement model of equation (1.1) the constant term α_1 provides for rigid body displacement. Third, the displacement models must include the constant strain states of the element. Similar to the second condition, this condition states that there should exist combinations of values of the generalized coordinates that cause all points on the element to experience the same strain. Displacement models of the type given in equation (1.1) when applied to 3D elasticity satisfy the three conditions given above, if at least the constant and linear terms are retained in the polynomials. In our finite element model, the displacement models used quadratic polynomials and are applied assuming 3D elasticity, therefore, the three convergence conditions are met.

The nodal displacements necessary to specify completely the deformation of the finite element are the degrees of freedom of the element. Each degree of freedom is specifically identified with a single nodal point and represents a displacement having a clear physical interpretation. The minimum number of degrees of freedom necessary for a given element is partly determined by the convergence requirements. The minimum number of nodal degrees of freedom was determined using a convergence analysis. In the analysis, the nodal degrees of freedom were increased until the nodal displacements tended to a solution (see Chapter 5 for results).

Due to the curves in the geometries of the femur and the tibia, the bones are very difficult to model using straight-sided elements such as the four-node tetrahedron. It is much more effective to model the bones using a higher order element, such as the isoparametric ten-node tetrahedron, as the curves can be better represented. In higher order elements, midside nodes are introduced to the primary nodes to match the number of degrees of freedom with the number of generalized coordinates in the displacement polynomial. Isoparametric elements use the same shape functions to define the element shape and displacement; the use of a quadratic shape function allows for the representation of curved boundaries. Moreover, fewer 10-node tetrahedral elements than

4-node tetrahedral elements are needed to achieve the same degree of accuracy in the final results.

Derivation of the element stiffness matrix using a variational principle

The selected element type and its associated displacement function are used to derive the stiffness matrix of the element. Specifically, the stiffness matrix for an element depends upon (1) the displacement model, (2) the geometry of the element, and (3) the local material properties or constitutive relations. Parameters such as the Young's modulus E and the Poisson's ratio ν can be used to define the local material properties. Since material properties are assigned to a particular finite element, inhomogeneity (e.g. for bone) can be accounted for by assigning different material properties to different finite elements in the assembly.

The derivation of element characteristic matrices and vectors involves the integration of the shape functions or their derivatives (or both) over the element. Three approaches can be used to derive the stiffness matrix: (1) the direct approach, (2) the variational approach, and (3) the weighted residual approach. The direct approach involves applying forces to an arbitrary element and calculating the appropriate element stiffness members from equilibrium equations. The variational approach involves applying energy or variational principles to form the stiffness matrices and force vectors; it is based upon minimizing the total energy in the system. In the weighted residual approach, the element matrices and vectors are derived directly from the governing differential equations of the problem. In solid mechanics, the variational approach and the weighted residual approach are the most common. In this study, the variational approach was used.

The variational approach applies the principle of minimum potential energy to obtain the stiffness matrix and load vector. The potential energy of an elastic body is defined as

$$\Pi = U + W_p \quad (1.7)$$

where Π is the potential energy, U is the strain energy, and W_p is the potential of the applied loads. We assume the forces remain constant during a variation of the displacements.

The principle of minimum potential energy is

$$\delta\Pi = \delta U + \delta W_p = \delta U - \delta W = 0 \quad (1.8)$$

where W is the work done on the system. The principle states that, of all possible displacement configurations a body can assume, the configuration satisfying equilibrium makes the potential energy assume a minimum value. The potential energy for a linearly elastic body can be expressed as the sum of the internal work (the strain energy due to internal stresses) and the potential of the body forces and surface tractions.

$$\Pi = \iiint_V dU(u, v, w) - \iiint_V (\bar{X}u + \bar{Y}v + \bar{Z}w)dV - \iint_{S_1} (\bar{T}_x u + \bar{T}_y v + \bar{T}_z w) dS_1 \quad (1.9)$$

where $dU(u, v, w)$ is the strain energy per unit volume (or strain energy density); $\bar{X}, \bar{Y}, \bar{Z}$ are quantified body forces; $\bar{T}_x, \bar{T}_y, \bar{T}_z$ are quantified surface tractions; S_1 is the surface of the body on which surface tractions are prescribed; and V is the volume of the body.

The strain energy density for a linear elastic body is defined as

$$dU = \frac{1}{2} \{\epsilon\}^T \{\sigma\} dV = \frac{1}{2} \{\epsilon\}^T [C] \{\epsilon\} dV \quad (1.10)$$

where $\{\epsilon\}$ is the vector of strains at any point within the element, $\{\sigma\}$ is the vector of stresses at any point in the element, and $[C]$ is the stress-strain matrix.

The total potential then becomes

$$\Pi = \frac{1}{2} \iiint_V (\{\epsilon\}^T [C] \{\epsilon\} - 2\{u\}^T \{\bar{X}\}) dV - \iint_{S_1} \{u\}^T \{\bar{T}\} dS_1 \quad (1.11)$$

where

$$\begin{aligned} \{u\}^T &= [u \quad v \quad w] \\ \{\bar{X}\}^T &= [\bar{X} \quad \bar{Y} \quad \bar{Z}] \\ \{\bar{T}\}^T &= [\bar{T}_x \quad \bar{T}_y \quad \bar{T}_z] \end{aligned}$$

Using interpolation functions, the displacements can be given as

$$\{u\} = [N] \{q\} \quad (1.12)$$

the strains can be given as

$$\{\varepsilon\} = [B]\{q\} \quad (1.13)$$

and the stresses can be given as

$$\{\sigma\} = [C][B]\{q\} \quad (1.14)$$

where $[B]$ is the strain-displacement matrix for interpolation models. Substituting these equations into equation (1.11), we can write the total potential in terms of the displacement model,

$$\Pi = \frac{1}{2} \iiint_V (\{q\}^T [B]^T [C] [B] \{q\} - 2\{q\}^T [N]^T \{\bar{X}\}) dV - \iint_{S_1} \{q\}^T [N]^T \{\bar{T}\} dS_1 \quad (1.15)$$

Applying the variational principle (equation (1.8)), we get

$$\{\delta q\}^T \left(\iiint_V [B]^T [C] [B] dV \{q\} - \iiint_V [N]^T \{\bar{X}\} dV - \iint_{S_1} [N]^T \{\bar{T}\} dS_1 \right) = 0 \quad (1.16)$$

Since the variations in the nodal displacements $\{\delta q\}$ are arbitrary, the expression in parentheses must be zero. This expression has the form

$$[k]\{q\} = \{Q\} \quad (1.17)$$

where the stiffness matrix and the nodal load vector are defined as

$$[k] = \iiint_V [B]^T [C] [B] dV \quad (1.18)$$

$$\{Q\} = \iiint_V [N]^T \{\bar{X}\} dV + \iint_{S_1} [N]^T \{\bar{T}\} dS_1 \quad (1.19)$$

Using the principle of minimum potential energy, the equilibrium equations for each individual element in the continuum can be obtained (Desai and Abel, 1972; Rao, 1982). In this study, the equilibrium equations for all elements were obtained with the aid of ANSYS[®]. In order for a solution to be obtained for the entire body, the individual equations must be combined together. The assembly process is described below.

Assembly of the algebraic equations for the overall discretized continuum

The equilibrium equations for the entire solution domain are obtained by combining the equations for the individual elements. The global stiffness matrix is

assembled from the individual element stiffness matrices; the global load vector is assembled from the element nodal force vectors. The direct stiffness method is employed in ANSYS® for assembling the algebraic equations in finite element applications partly because of the ease of coding the technique for the computer.

The basis for an assembly method is that the nodal interconnections require the displacements at a node to be the same for all elements which include that node. The procedure of assembling the element matrices and vectors is based on the requirement of compatibility at the element nodes. This means that the nodes where elements are connected, the values of the unknown nodal degrees of freedom are the same for all the elements joining at that node.

Solutions for the unknown displacements

The algebraic equations assembled in above step are then solved for the unknown displacements. In linear equilibrium problems such as the case in this study, a straightforward application of matrix algebra techniques is used.

There are two basic approaches for the solution of large systems of equations: elimination and iteration. The elimination approach, typified by Gaussian elimination, is a procedure in which the stiffness matrix is transformed to a triangular form to be solved directly for the unknowns. An extension of the elimination approach is the frontal processing method, which solves and eliminates as the matrix is being formed element by element. In this study, the computer must handle extremely large numbers of equations, which requires a large computer core space. As computer core space was not sufficient to handle the storage of the extremely large system of equations, an iterative approach to solving the system was used.

Iterative methods start with an initial approximation and, by applying a suitably chosen algorithm, lead to successively better approximations. When the process converges, we can expect a good approximate solution. The main advantages of iterative methods are the simplicity and uniformity of the operations to be performed (making them well-suited for computer use), and their relative insensitivity to growth of round-off errors. The conjugate gradient method, which is based on the principle of unconstrained

minimization of a function, was used to solve for the unknown displacements in this work.

Computation of the element strains and stress from the nodal displacements

In this study, the distribution of displacements and stresses is of interest. The displacements were obtained using an iterative solver in ANSYS® as described above. By applying constitutive relations such as equations (1.13) and (1.14), the distribution of stresses and strains can be determined. In this study, the post-processor in ANSYS® aided in the determination of the internal stresses and strains of the bones.

The material properties of bone depend on orientation, making it an anisotropic material. It has been shown that bone is an orthotropic material; having three planes of symmetry. For linearly elastic orthotropic 3D solids, the stress-strain relations are given by Hooke's law.

The state of stress in an elemental volume can be expressed in vector form as

$$\{\sigma\} = [\sigma_x \quad \sigma_y \quad \sigma_z \quad \tau_{xy} \quad \tau_{yz} \quad \tau_{zx}] \quad (1.20)$$

where σ_x , σ_y , and σ_z are the normal components of stress, and τ_{xy} , τ_{yz} , and τ_{zx} are the components of shear stress. Corresponding to the six stress components, the state of strain at a point can be divided into six strain components

$$\{\varepsilon\} = [\varepsilon_x \quad \varepsilon_y \quad \varepsilon_z \quad \gamma_{xy} \quad \gamma_{yz} \quad \gamma_{zx}] \quad (1.21)$$

The relations between strain and displacement components u , v , and w at a point can be simplified as

$$\begin{aligned}
\varepsilon_x &= \frac{\partial u}{\partial x} & \gamma_{xy} &= \frac{\partial v}{\partial x} + \frac{\partial u}{\partial y} \\
\varepsilon_y &= \frac{\partial v}{\partial y} & \gamma_{yz} &= \frac{\partial w}{\partial y} + \frac{\partial v}{\partial z} \\
\varepsilon_z &= \frac{\partial w}{\partial z} & \gamma_{zx} &= \frac{\partial u}{\partial z} + \frac{\partial w}{\partial x}
\end{aligned} \tag{1.22}$$

As an extension of Hooke's law, each of the six stress components can be expressed as a linear function of the six strain components

$$\begin{Bmatrix} \sigma_x \\ \sigma_y \\ \sigma_z \\ \tau_{xy} \\ \tau_{yz} \\ \tau_{zx} \end{Bmatrix} = \begin{bmatrix} C_{11} & C_{12} & \dots & C_{16} \\ C_{21} & C_{22} & \dots & C_{26} \\ \cdot & & & \cdot \\ \cdot & & & \cdot \\ \cdot & & & \cdot \\ C_{61} & C_{62} & \dots & C_{66} \end{bmatrix} \begin{Bmatrix} \varepsilon_x \\ \varepsilon_y \\ \varepsilon_z \\ \gamma_{xy} \\ \gamma_{yz} \\ \gamma_{zx} \end{Bmatrix} \tag{1.23}$$

or

$$\{\sigma\} = [C]\{\varepsilon\} \tag{1.24}$$

The strain vector can be related to a displacement model using a strain-displacement matrix

$$\{\varepsilon\} = [B]\{q\} \tag{1.14}$$

The stress vector can then be related to the nodal displacement vector via an appropriate stress-strain matrix

$$\{\sigma\} = [C][B]\{q\} \tag{1.15}$$

Chapter 2: Literature Review of Knee-Related Finite Element Applications

This chapter provides a brief overview of knee-related FEA applications. The analysis of the state of finite element modeling pertaining to knee biomechanics will be discussed on two levels, the stress states at the macrostructural and microstructural levels. The macrostructure level focuses on the development of the tibiofemoral joint as a complete structure, while the microstructural level discusses the modeling of trabecular bone.

2.1 Finite Element Analysis in Knee Biomechanics

The finite element method (FEM) was introduced in 1972 as a tool for evaluating complex problems in orthopaedic biomechanics (Brekelmans et al., 1972). Since then, the FEM has provided a solid mechanics aspect to the discussion of the etiology of diseases of joint such as the hip, e.g. osteonecrosis (Brown and Hild, 1983) and Perthes' disease (Ueo et al., 1985), and the knee, e.g. osteochondrosis dissecans (Nambu et al., 1991) and lesions in degenerative arthritis (Hayes et al., 1978). The most extensive application of the FEM to tibiofemoral biomechanics, however, has been in artificial joint design and fixation. With the exception of a study by Pissinou and Brook (1980) who looked at femoral and tibial prostheses components, nearly all studies have dealt with the tibial component fixation. This is not surprising as loosening of the tibial component was reported in 60% of replaced knees (Insall et al., 1976) with the most common site of loosening at the bone-cement interface. The FEM has provided insight into the origins of the loosening; consequently, many fixation aspects of tibial prostheses have been individually examined using FE models. These aspects include the design of the prosthesis itself (e.g. trays, posts, screws) and the fixation of the prosthesis (e.g. bone cement thickness, cement type).

Three aspects of the tibial tray have been investigated: the type, thickness, and width of the tray. It was found that metal trays reduced compression stresses compared

to polyethylene trays in both bone (Murase et al., 1983; Bartel et al., 1980; Askew et al., 1978; 1980) and bone cement (Askew et al., 1978, 1980; Walker et al., 1981). The width of the tray was reported to have very little effect on stress distributions, with the exception being the seldom-used wide metal-backed polyethylene trays (Murase et al., 1983); the thickness of the tray was also reported to have little effect on stress distributions (Shrivastava et al., 1980).

The addition of a central fixation post in tibial prostheses was found to have a positive effect by reducing stress levels in bone cement (Askew et al., 1978, 1980; Walker et al., 1981) and the cancellous bone surrounding the post (Askew et al., 1978, 1980). The stress level is particularly affected by the post length. A short post has little effect on the stresses (Murase et al., 1983; Bartel et al., 1980) while a longer post reduces stress in the bone beneath the tibial tray in both compression (Askew and Lewis, 1981; Lewis et al., 1982; Murase et al., 1983) and tension (Lewis et al., 1982). However, when eccentrically loaded, a longer post will cause greater tension in the bone cement due to bending of the post (Eibeck et al., 1979). Similar to the tibial tray, a metal post produces less cement stresses than a polyethylene post (Lewis et al., 1982). Cementless fixation of the tray can also be achieved using cancellous screws and pegs. The use of these fixation screws was found to reduce micromotion at the bone-implant interface compared to pegs (Hashemi and Shirazi-Adl, 2000).

Fixation of the prosthesis to the bone is commonly done with polymethylmethacrylate (PMMA), also known as bone cement. Aseptic loosening of the tibial component is often seen at the bone-cement interface and may be caused by fracture of the bone cement. A few studies have isolated the cement itself for investigation with some contradictory results. While cement thickness was not found to affect bone stress distribution (Shrivastava et al., 1980) nor cement stress levels (Walker et al., 1981), Eibeck et al. (1979) has found that increasing cement thickness increases shear stresses in cement. However, the parameters in each of the models (e.g. geometry, material properties) were different, which may explain the varied results. An alternative to PMMA involves the pressure injection of a low-viscosity cement resulting in a cement-bone composite. The composite exhibits a stiffness that is between cement and trabecular bone. A comparison of the cement-bone composite with bone cement itself

showed that the composite allows for a more direct transfer of load to the cortex without unduly stressing trabecular bone (Shrivastava et al., 1982). FE prosthesis models have often assumed a perfect bond between the prosthesis, cement, and bone. A perfect bond assumes there is no relative displacement or material discontinuity at the cement-prosthesis and cement-bone interfaces. This may be an inaccurate representation as radiolucent lines and fibrous tissue has been found at the cement-bone interface suggests imperfect bonding or bond degeneration (Lewis et al., 1982). More accurate interface modeling can be achieved using nonlinear friction models to describe the interface as done by Hashemi and Shirazi-Adl (2000).

Additional work is still needed to develop finite element models as previously mentioned models (e.g. tibial prosthesis models) may not have a realistic representation of geometry, bone properties, or loading conditions due to the reasons discussed below. Finite element models of the knee vary in many aspects, e.g. geometry, material properties, and loading conditions. Many of the early finite element models were considerably simplified because of the prohibitive amount of manual labour needed to generate finite element models in addition to computational limitations. As a result, early FE tibial geometries were either two-dimensional (Eibeck et al., 1979; Askew and Lewis, 1981) or axisymmetric (Hayes et al., 1978; Murase et al., 1983; Vichnin et al., 1979; Van Campen et al., 1979; Shrivastava et al., 1982). Within the two-dimensional plane, most geometries were also symmetric along the longitudinal axis, further reducing the accuracy of the representation. Indeed, a few of the geometries showed no curvature whatsoever. Studies using these geometries were not quantitatively accurate but cost effective. The 2D and axisymmetric geometries were often employed for parametric studies where they allowed for a qualitative comparison. Evidently, the analyses in these studies were restricted to the mid-coronal and mid-sagittal planes. Another restriction of these models was the lack of convergence analysis. Only one study (Murase et al., 1983) reported a convergence analysis. The mesh size appears to have been arbitrarily chosen, with the resulting number of elements being less than 500. The appropriateness of the mesh size assigned was only evaluated in the model by Askew and Lewis (1981) by comparison with an analytical solution. These results showed good agreement when compared with a similar 3D FE model (Lewis et al., 1982), justifying that a simplified

model can sometimes be appropriately employed. However, a three-dimensional geometry can allow for a much more comprehensive analysis because of the asymmetric geometry of the femur and the tibia. Earlier 3D models (Nambu et al., 1991; Hashemi and Shirazi-Adl, 2000; Little et al., 1986; Lewis et al., 1982) did not have automatic mesh generation available and still required a large amount of time to generate meshes. The meshing appeared to be very coarse and eight-node brick elements were mostly used. These two factors may contribute to an inaccurate representation of the many curves in the femur and tibial surfaces. Inaccurate representation of the geometry will lead to inaccurate representation of the stress and strain states (Huiskes et al., 1981; Rohlmann et al., 1982). The numerical results were validated in only a few studies (Van Campen et al., 1979; Shrivastava et al., 1982; Little et al., 1986) and only general trends in the strains matched. Thus, the above studies show that finite element models may be used qualitatively for parametric studies of prostheses, they cannot be applied quantitatively (and therefore clinically) without a prohibitive amount of caution. It is expected that inaccurate geometry may contribute to a lack of quantitative agreement; however, it is more likely that inappropriate application of material properties to the FE models contributes a greater factor to this lack of agreement.

Incorporating realistic representations of bone properties is necessary to create accurate FE models (Huiskes et al., 1981; Rohlmann et al., 1982). There is general agreement that bone properties incorporated into FE models can be assumed to be linearly elastic (Cowin, 1989). Non-prosthesis FE models have incorporated articular cartilage, subchondral bone, cortical bone, and cancellous bone. Hayes et al. (1978) have found that articular cartilage has less of an effect in transmitting loads than the bones. Large loads go through articular cartilage undiminished to subchondral bone, which transmits the loads to the diaphyseal cortex. As such, the lack of a subchondral bone layer in the femur model of Nambu et al. (1991) may affect their strain results, particularly since the region of interest is directly beneath the subchondral layer. In addition to this, neither the heterogeneity nor the anisotropy of the bones were considered.

Both cortical and cancellous bones have been shown to exhibit spatial variation in the femur and tibia (Rho, 1992). With the exception of a few studies using FE models

(Nambu et al., 1991; Shrivastava et al., 1982), the heterogeneity of cortical and cancellous bones has been considered. Different methods have been employed to incorporate heterogeneity, including using bone-area fractions (Askew and Lewis, 1981; Lewis et al., 1982), mapping experimental data (Hashemi and Shirazi-Adl, 2000; Murase et al., 1983; Hayes et al., 1978; Little et al., 1986), and using CT scanning (Keyak et al., 1990; Wirtz et al., 2003), with varying degrees of heterogeneity. The spatial variation of the properties of cancellous bone is particularly important as it promotes the transfer of stresses to the cortex (Askew and Lewis, 1981). As such, the accuracy of stress distributions in early FE studies (which lacked sufficiently detailed experimentally mapped data) may have been compromised (e.g. the models by Askew and Lewis (1981) and Lewis et al. (1982) only incorporated 4 different cancellous bones). Heterogeneity in cortical bone is not as great as in cancellous bone (Rho, 1992). However, the variation should still be accounted for as different cortical stiffness has been shown to affect peak stresses (Vichnin et al., 1979) in the cortex as well as stresses beneath the tibial tray in prostheses (Askew and Lewis, 1981). To the author's knowledge, a three-dimensional FE model of the distal femur or proximal tibia has not been developed that has incorporated both cortical and cancellous bone heterogeneity.

A second characteristic of bone that needs to be adequately implemented into FE models is anisotropy. Anisotropy of cortical bone is more frequently modeled than cancellous bone. However, cancellous bone is much more anisotropic than cortical bone (Rho, 1992). Few FE models (e.g. Askew and Lewis (1981) and Van Campen et al. (1979)) have considered the anisotropy of cancellous bone. Rather than incorporating both heterogeneity and anisotropy, many FE models have only considered heterogeneity, often citing that heterogeneity is more important than anisotropy and, in itself, is sufficient to model the behaviour of bone. Askew and Lewis (1981) have previously stated that anisotropy affects proximal post stress behaviour in TKR FE models. However, in a later study (Lewis et al., 1982), the same group stated that, because the exact nature of anisotropy of trabecular bone was not known, the inclusion of anisotropy was not warranted (the assumption of heterogeneity would be sufficient). Huiskes et al. (1981) also agreed that heterogeneity alone can accurately characterize bone, if the loading on the bone was mainly axial. As loading in the majority of these FE studies was

purely axial, it may have seemed that ignoring anisotropy was adequate. However, in the study by Little et al. (1986) in which heterogeneity was extensively (and exclusively) modeled, there were marked differences in strains when the FE model was compared with *in vitro* experimental data. It is unlikely the discrepancies are due purely to the exclusion of anisotropy, however, there is strong possibility that results could be improved with its inclusion. The inconsistency between the finite element solutions and experimental data limit the applications of reported FE models. To develop more accurate FE models, more experimental data mapping the anisotropy of cancellous bone of the femur and the tibia need to be made available. Only a few studies have published maps of bone stiffness (Goldstein et al., 1983; Rho, 1992, 1996; Ashman et al., 1989); to the author's knowledge, no studies have reported the mapping of shear modulus or Poisson's ratio. Recently, Wirtz et al. (2003) developed an orthotropic FE model of the proximal femur, but acknowledged the model requires accuracy analysis and validation. To the author's best knowledge, no three-dimensional FE models of the distal femur or the proximal tibia have been developed that have considered both the anisotropy (orthotropy) and inhomogeneity of the mechanical properties of bone.

Physiological loading is an important aspect of FE modeling which has not been accurately represented in past models. It is generally accepted that an appropriate physiological load in the tibiofemoral joint is approximately 3 times body weight as found by Morrison (1970) and Harrington (1976). The distribution of this force is not evenly distributed between the medial and lateral condyles, but rather distributed with a ratio of 55:45 (medial to lateral). However, Little et al. (1986) found the effects of this distribution to be minor. The majority of FE models have employed simplified loading conditions of one- and two-point loading (Eibeck et al., 1979; Lewis et al., 1982; Askew and Lewis, 1981), Fourier cosine and sine series expansion (Hayes et al., 1978; Murase et al., 1983; Van Campen et al., 1979; Shrivastava et al., 1982), or uniform load distribution (Nambu et al., 1991; Hashemi and Shirazi-Adl, 2000). Fourier series expansion allows for a non-uniformly distributed loading and is used primarily in axisymmetric FE models. Point loading is used to represent extreme functional performance loads and is considered realistic in these extreme cases (Lewis et al., 1982). In parametric failure studies of prosthetic designs, point loading is appropriate. However, point loading will somewhat

enhance stresses beneath the load points (Lewis et al., 1982), and reduce the accuracy of resulting stress levels and distributions. A more suitable case for point loading is for hinged tibial prostheses in which line forces are transferred along the hinge axis (Eibeck et al., 1979). Contact patterns in the knee joint have been experimentally determined (Walker and Hajek, 1972; Fukubayashi and Kurosawa, 1980) and should be implemented into FE models to increase the accuracy of the results. Early 3D models could not accurately represent these patterns due to the coarseness of the mesh. Nambu et al. (1991) attempted to model the elliptical contact patterns found by Walker and Hajek (1972) but, due to low mesh density, the nodal loading produced a pattern appearing more trapezoidal than elliptical. However, more recent 3D models with greater mesh density still have not implemented physiologically realistic loading. Hashemi and Shirazi-Adl (2000) used uniform loading in a rectangular contact area while Little et al. (1986) implemented a non-uniform distributed loading with a trapezoidal pattern. In this study, a method has been developed to attempt to address the deficiencies in physiological loading seen in other FE models.

In addition to accurate loading of the joint, adequate constraints are also needed at the proximal femur and distal tibia. In all the FE models discussed above, the diaphysis of the bones is fully constrained. Nambu et al. (1991) performed a sensitivity analysis on the effect of fully constraining the base and found it had little effect on strains and displacements as long as the region of interest was far away from the base. In this study, the region of interest is at the proximal tibia and distal femur. A short investigation into the effect of fully constraining the base on the bone stresses was performed to determine if these boundary conditions were adequate.

All the FE models reported above contained less than 10,000 nodes. For qualitative parametric studies, it may not be completely necessary to implement highly accurate representations of geometry, material properties, and loading conditions (Shrivastava et al., 1982). However, the numerical results of these parametric models are often not validated and are therefore limited in their clinical applicability. The model developed in this study addresses to a great extent the deficiencies of previous FE studies. This was done for both the femur and tibia, which has not been seen in majority of previous works.

2.2 FEA in Microstructural Analysis of Bone

The modeling of bone at the microstructural level is beyond the scope of this work. However, a very brief discussion is provided to demonstrate to the reader the potential of the finite element method. As a result of advances in technology, the use of the finite element method is becoming increasingly prominent in the stress analysis of cancellous bone at the microstructural level. One of the earliest studies used the FEM in combination with experimental methods to determine the anisotropic properties of cancellous bone by modeling two-dimensional trabecular bone (Williams and Lewis, 1982). Individual trabeculae were modeled by Mente and Lewis (1989) to determine the average elastic modulus of trabecular tissue. Many of the early studies of trabecular bone were limited by their accuracy and applicability. Traditional testing on single trabeculae was difficult because of its small size; models of trabecular bone were simplified by the assumption of a repetitive structure. A new method using micro-CT scanning to create finite element models was developed by van Rietbergen et al. (1995), which is now frequently adopted for trabecular bone studies. More recently, this technique has been used to examine the changes in stiffness of subchondral trabecular bone due to osteoarthritis (Day et al., 2001). Finite element models are currently being used to help determine the properties that can fully characterize trabecular bone stiffness. These properties include mineral distribution (Van der Linden et al., 2001), volume fraction and fabric (Kabel et al., 1999; van Rietbergen et al., 1998). The computational approach using the FEM is also being used to study the influence of mechanical factors on bone remodeling at the trabecular level to help understand the relationships between loading conditions on trabecular bone (Adachi et al., 2001), and functional adaptation (Tsubota et al., 2002). A future application of the full characterization of trabecular bone stiffness is the incorporation of this information into our FE models, allowing for a more accurate analysis of the stress states in cancellous bone as applied to knee reconstructions and total knee replacements.

Chapter 3: Geometry of Finite Element Model

3.1 Creation of surfaces (CAD modeling)

Composite femora and tibiae are widely used as mechanical surrogates in orthopaedic research. These synthetic models offer the advantage of better availability and less variability than cadaveric specimens. The commercially available composites (Sawbones, Pacific Research Labs, Vashon Island, WA) consist of glass fiber reinforced epoxy (representing cortical bone) and polyurethane foam (representing cancellous bone). Experimental validation of the second-generation composite femur and tibia (Cristofolini et al., 1996; Cristofolini and Viceconti, 2000), and third-generation composite femur and tibia (Heiner and Brown, 2001) has shown these synthetic models to be acceptable substitutes for certain types of testing.

In 1996, a computational analogue to the second-generation composite femur was introduced called the Standardized Femur (Viceconti et al., 1996). The model, created by the Istituti Ortopedici Rizzoli, was based on CT scans of the composite femur. The availability of the model in the public domain was intended to eliminate the time-consuming step of generating a model and lend itself to comparative studies.

Third-generation composite models from Sawbones (Pacific Research Labs, Vashon Island, WA) allowed for greater anatomic detail, including oval shaped diaphyseal cross-sections and the addition of a linea aspera. The geometries of the composite bones were based on left side cadaver femur and tibia bones from an 890N (90.8 kgf), 183cm tall male. The computational analogues to these newer composite models were created by Greer (1999), and are available on the Internet at the International Society of Biomechanics Finite Element Repository managed by the Istituti Ortopedici Rizzoli, Bologna, Italy (ISB, 2001). The surface geometries were created from laser-scanned data. A Hymarc laser scanner from 3-D Vision Systems (Nepean, ON) was used for the complete scanning of the femur, and Imagineware (SDRC, Ann Arbor, MI) was used to extrapolate the curve geometry. The curves were exported as IGES files and the solid model was created with Pro/Engineer® (Parametric Technology Corp., Waltham, MA). The completed femur and tibia models can be seen in Figs 3.1

and 3.2, respectively. The solid models were exported to ANSYS® (Swanson Inc., Houston, PA) on an Athlon XP1800+ PC via IGES neutral format. The models were imported into ANSYS® as a defeatured model, which allowed automatic volume creation. The geometries obtained from the ISB are strictly surface models containing only geometrical information and no material properties or meshing were associated with them.

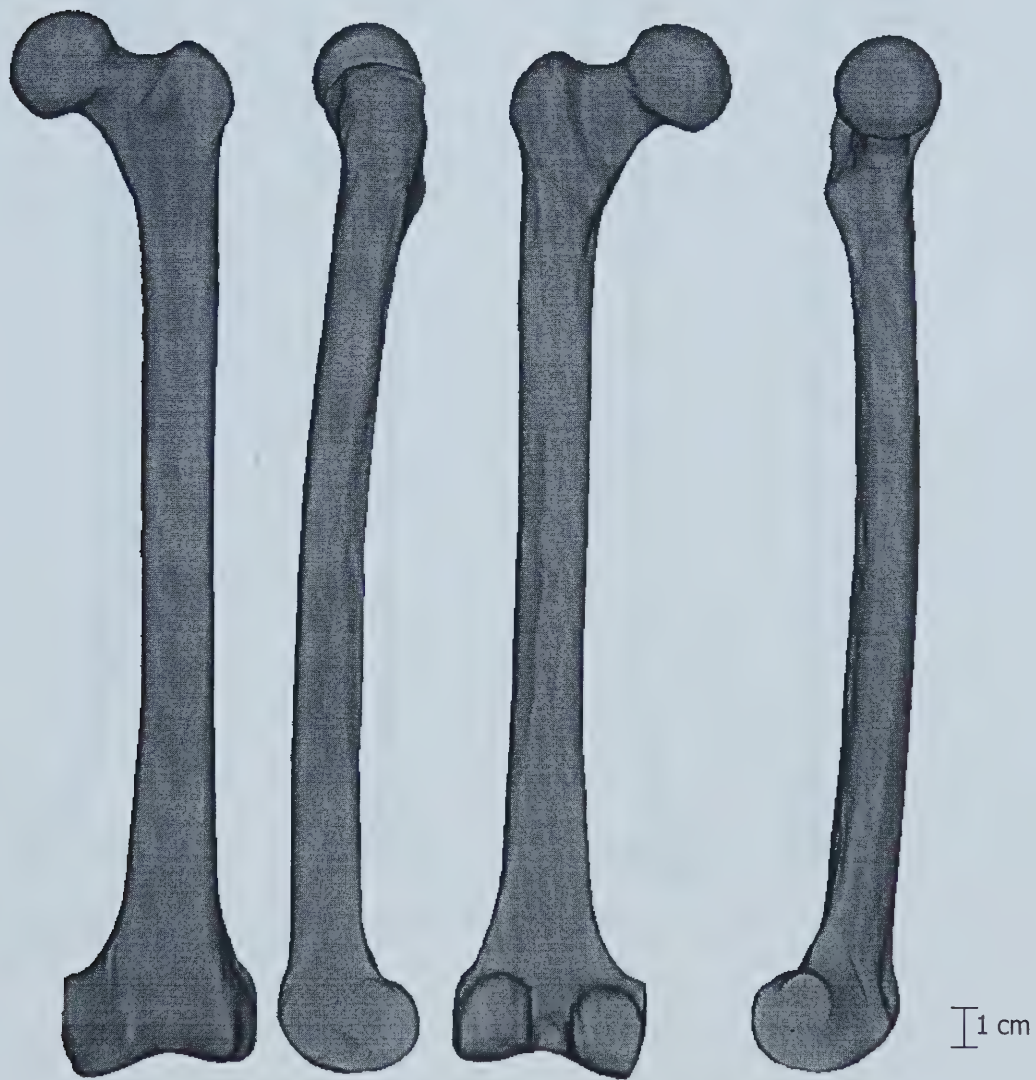


Figure 3.1: From left to right: anterior, lateral, posterior, and medial views of the femur rendered in Pro/Engineer®.

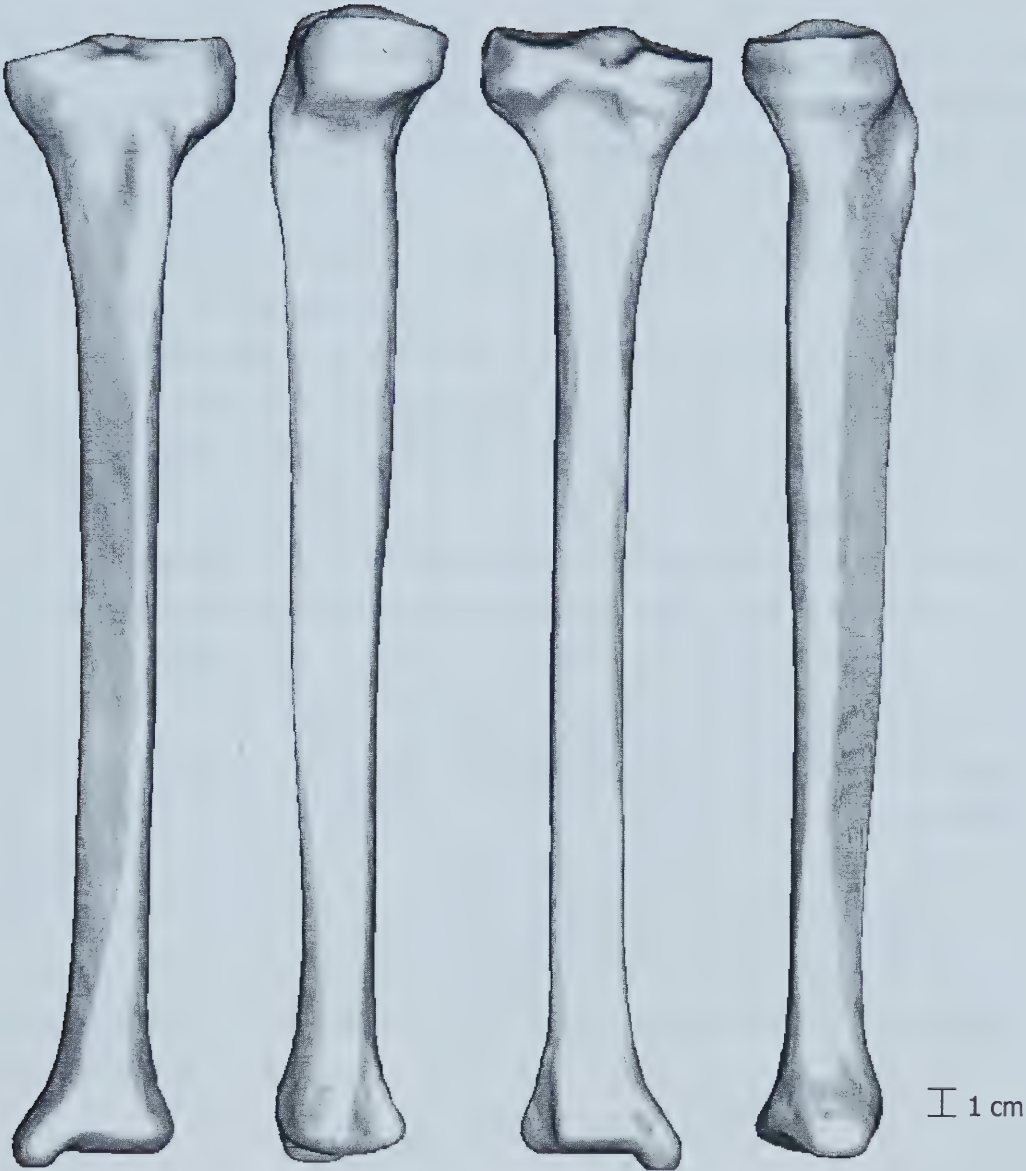


Figure 3.2: From left to right: anterior, lateral, posterior, and medial views of the tibia rendered in Pro/Engineer®.

3.2 Sensitivity Analysis

As many simulations would be run using the femur and tibia models, it was necessary to optimize the computational effort without compromising the accuracy of these simulations. A reduction of the computational time could be achieved by modeling the immediate region around the knee joint, i.e. the distal femur and the proximal tibia. By meshing a reduced section of the femur and tibia, the number of nodes would be reduced along with computational time.

Sensitivity analyses of the femur and tibia FE models were performed to determine the optimal truncation lengths for each of the models. The stress levels were monitored at five locations on the tibia (Fig 3.3) and at five locations on the femur (Fig 3.4). The locations were selected, firstly, because they were in regions of interest in future stress analyses. Secondly, to allow for the best comparison possible, the ten nodes were selected because they had identical coordinates across all the mesh densities. All the monitored points were on the cortical surface. It was assumed that the cortical surface would be more sensitive to the change in length, therefore no cancellous bone points were monitored. Anisotropic and inhomogeneous bone properties were assigned (see Chapter 4 for further elaboration). The models were meshed with tetrahedral elements of 2 mm (see Chapter 5 for further discussion). Loading of 1000 N of compression was used, and the truncated ends of the models were rigidly fixed.

Von Mises stresses were used to evaluate the effect of the changing lengths. It represents a nominal stress magnitude and, in a three-dimensional state, von Mises stress (σ_{VM}) is defined as

$$\sigma_{VM} = \frac{1}{\sqrt{2}} \sqrt{(\sigma_1 - \sigma_2)^2 + (\sigma_2 - \sigma_3)^2 + (\sigma_3 - \sigma_1)^2} \quad (3.1)$$

where σ_1 , σ_2 , and σ_3 are the three principal stresses.

Strictly speaking, von Mises stress may not be the most suitable measure of mechanical stress in anisotropic, heterogeneous materials such as bone. A combination of principal, von Mises, and normal stresses can be used to gain a better sense of the stress state. However, von Mises stress is a relatively convenient, scalar measure of

stress at a point. Von Mises stress is used in this study because it allows for a simple comparison of stresses among the different FE models generated.

Von Mises stress at each of the five points was plotted as a function of the total length of the tibia (Fig 3.5). Most of the stress values at the monitored points showed little sensitivity to the diaphyseal lengths. This analysis indicated that a total tibia length of approximately 80 mm would be adequate. A similar analysis was repeated to determine an adequate truncation length for the femur. Von Mises stresses at five points on the cortex of the femur were plotted as a function of the total length of the femur (Fig 3.6). The stresses showed relatively little sensitivity to the diaphyseal lengths after an approximate length of 90 mm. Therefore, a total femur length of approximately 90 mm was assumed to be adequate.

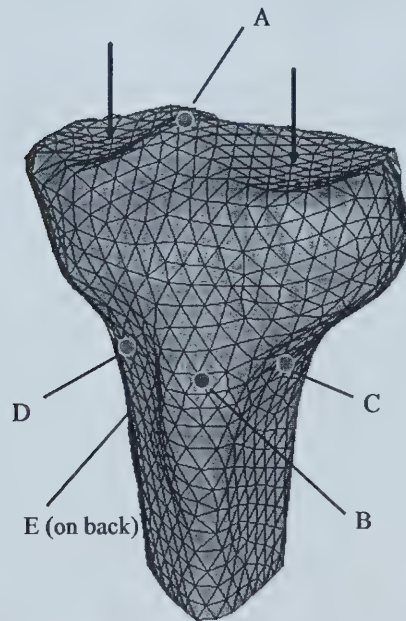


Figure 3.3: Points A to E are the five stress observation locations on the tibia FE model used in the sensitivity analysis. The arrows represent the loads placed on the model.

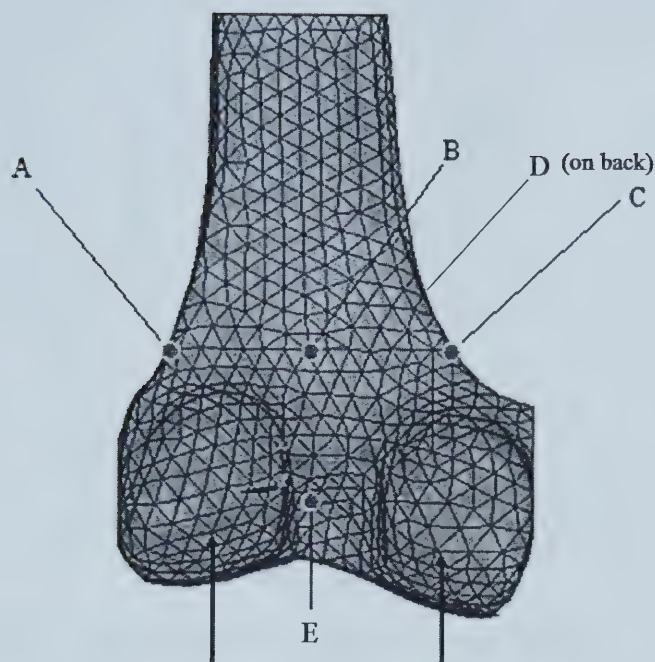


Figure 3.4: Points A to E are the five stress observation locations on the femur FE model used in the sensitivity analysis. The arrows represent the loads placed on the model.

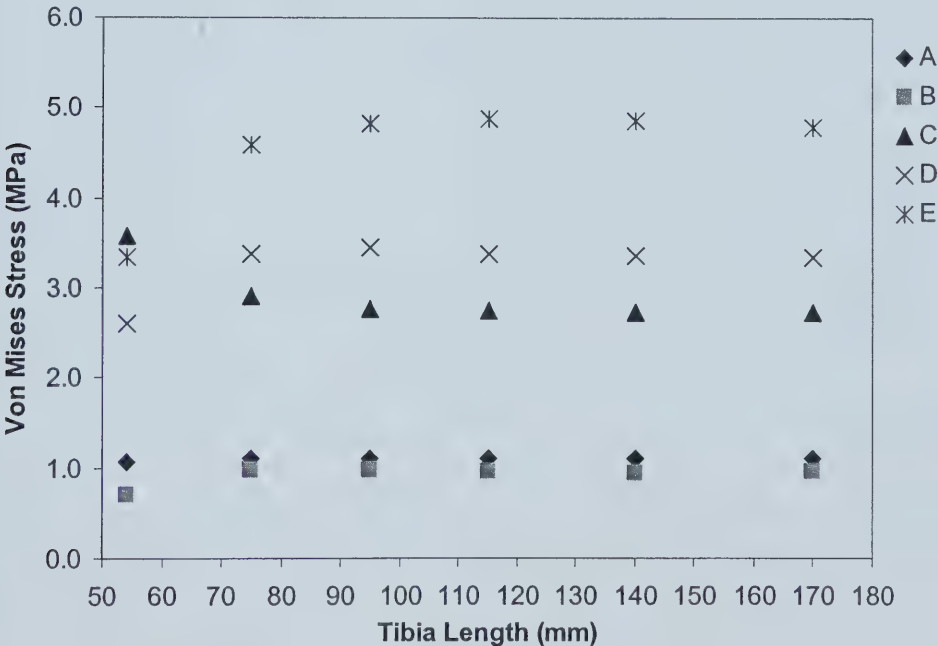


Figure 3.5: Von Mises stress of five monitored points, A to E (see Fig 3.3), on tibia versus the total length of the tibia.

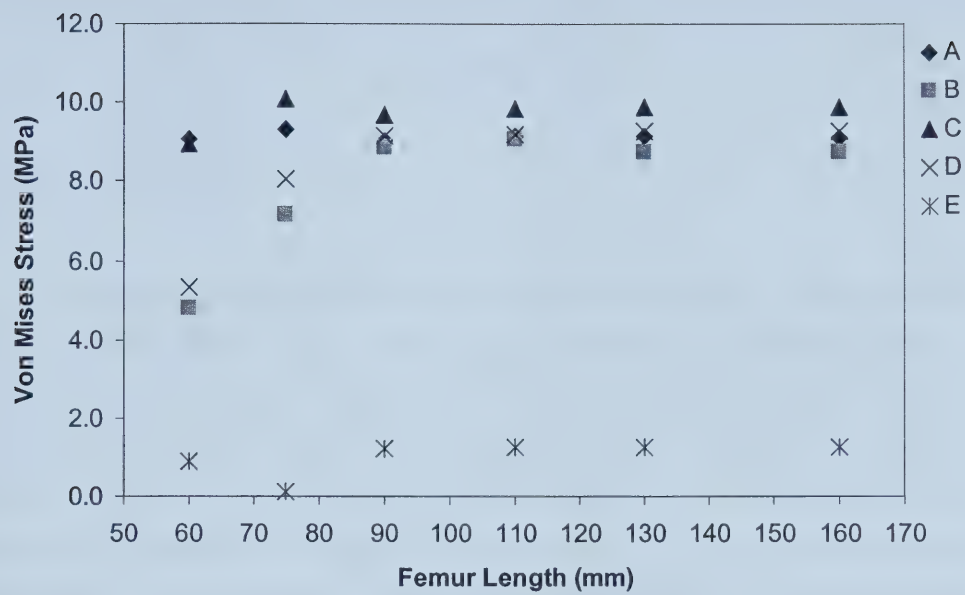


Figure 3.6: Von Mises stress of five monitored points, A to E (see Fig 3.4), on femur versus the total length of the femur.

Chapter 4: Selection of Material Properties of Finite Element Model

4.1 Background

The bone in the femur and tibia is composed of three types: cortical, cancellous, and subchondral (Figure 4.1). Cortical bone comprises an outer dense shell and cancellous bone comprises the spongy inner bone tissue. Cortical bone is stronger than cancellous bone and protects the softer cancellous bone from damage. The subchondral bone is a thin layer of bone beneath the articular cartilage. Many authors have examined the mechanical properties of cortical and cancellous bone, and a large range of values has been reported. In this chapter, some of the published data will be reported and a discussion of the selection of the material properties implemented in the FE models will be presented.

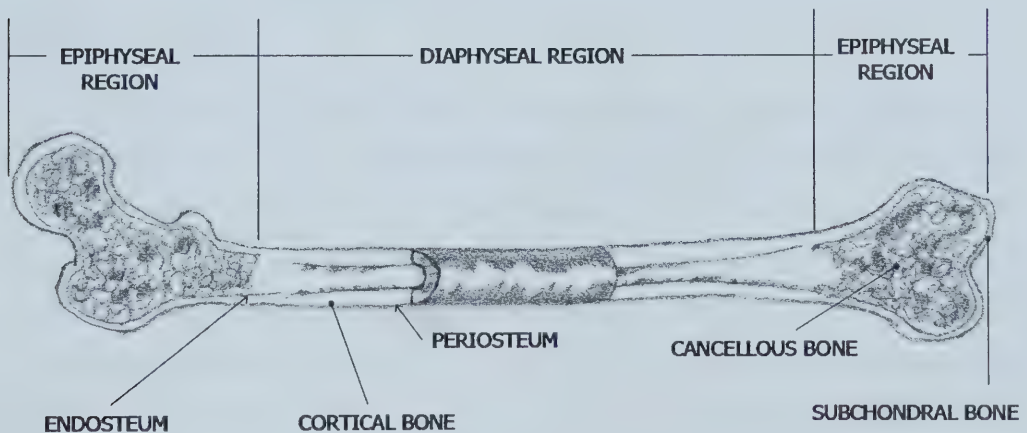


Figure 4.1: Longitudinal section of the human femur, illustrating cortical and cancellous bone types.

4.1.1 Cortical Bone

Cortical bone tissue is a complex biological matrix. All normal adult human cortical bone consists of a region of secondary osteons (or secondary Haversian bone)

bounded on the endosteal and periosteal surfaces by circumferential lamellar bone (Figure 4.2). Lamellar bone is characterized by many sheets of bone separated by thin interlamellar cement bands. It is present in both secondary osteons and circumferential lamellar bone. Each sheet of lamella exhibits a predominant fiber orientation that may be either identical or different in successive sheets (Carter and Spengler, 1978).

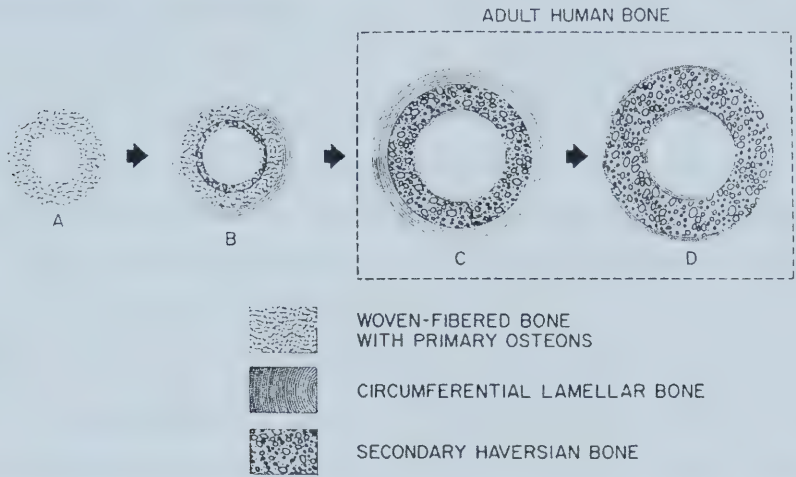


Figure 4.2: Maturation of human cortical bone. Reproduced from Carter and Spengler (1978) with permission.

Cortical bone has been found to exhibit anisotropic properties, specifically it is stronger and stiffer in the longitudinal direction (parallel to the long axis of the bone) than in the transverse direction (Carter and Spengler, 1978). However, it is not certain whether cortical bone should be characterized as a material that is transversely isotropic or orthotropic. A transversely isotropic material displays different material properties in two perpendicular directions, and requires only five independent constants: two moduli of elasticity, two Poisson's ratios, and a shear modulus. An orthotropic material displays different material properties in three mutually perpendicular directions, and requires nine independent constants: three moduli of elasticity, three shear moduli, and three Poisson's ratios.

Column 1 in Table 4.1 displays the reported stiffness of femoral cortical bone found by Reilly and Burstein (1975). The bone used in the study was obtained from the middle third of cadaver bone ranging in age from 19 to 80 years old. Mechanical tension and compression tests were used to obtain the data. Reilly and Burstein (1975)

postulated that the large standard deviations in the constants (7 – 10% for Young's moduli; 30% for Poisson's ratios) are likely due to the inhomogeneous nature of the bone material and the small specimen size (5 x 5 x 25 mm). In addition, they state the high Poisson's ratios show that cortical bone is not so much a material but more of a structure with a high degree of kinematic deformation.

The stiffness of cortical bone was also studied in a review by Van Buskirk and Ashman (1981). The review presented data from a Latvian group (Knets and Malmeisters, 1975) who tested bone in six areas around the mid-diaphysis region of the human tibia. The Latvian group found significant differences in nearly all the stiffness coefficients in each of the six areas. The mean results are shown in the second column of Table 4.1. When compared with the data from Reilly and Burstein (1975), the E_1 and E_2 values found by Knets and Malmeisters (1975) are lower. Van Buskirk and Ashman (1981) felt this could be due to small specimen sizes and end effects. When small cubed specimens are compressed on a testing machine, its sides must move outward in a direction perpendicular to the direction of the load (Reilly and Burstein, 1974). As the load increases, however, friction limits this movement at the sites of specimen-machine contact and the specimen under load becomes barrel-shaped. This so-called edge effect is accentuated when the test specimens are small (Reilly and Burstein, 1974) and might have contaminated the data from the Latvian group. However, the differences between the two sets of data may also arise from the inherent difference in stiffness between femoral and tibial cortical bones (Burstein et al., 1976).

A method to avoid edge effects is to use ultrasound to test material properties. The accuracy of ultrasound testing has been confirmed to be comparable to mechanical testing (Ashman et al., 1989). Using ultrasound, Van Buskirk and Ashman (1981) found orthotropy in cortical bone and confirmed the significant differences in the stiffness coefficients found by Knets and Malmeisters (1975). The mean data are presented in column three of Table 4.1. Further studies using ultrasound repeatedly showed the stiffness of cortical bone to be significantly different in three mutually perpendicular directions (Rho, 1992, 1996; Ashman et al., 1989). The data of Ashman et al. (1984) and Rho (1992) are shown in columns four and five, respectively, in Table 4.1.

Table 4.1: Mechanical properties of human cortical bone published in literature. Directions 1 and 2 are the radial and circumferential directions, respectively. Direction 3 is coincident with the long axis of the bone.

Studies	Reilly and Burstein (1975)	Knets and Malmeisters (1975)	Van Buskirk and Ashman (1981)	Ashman et al. (1984)	Rho (1992)
Bone	Femur	Tibia	Femur	Femur	Tibia
E_1	11.5	6.91	13.0	12.0	11.7
E_2	11.5	8.51	14.4	13.4	12.2
E_3	17.0	18.4	21.5	20.0	20.7
G_{12}	--	2.41	4.74	4.53	4.1
G_{13}	3.28	3.56	5.85	5.61	5.17
G_{23}	3.28	4.91	6.56	6.23	5.7
ν_{12}	0.58	0.488	0.37	0.376	0.420
ν_{13}	--	0.119	0.24	0.222	0.237
ν_{23}	--	0.142	0.22	0.235	0.231
ν_{21}	0.58	0.622	0.42	0.422	0.435
ν_{31}	0.46	0.315	0.40	0.371	0.417
ν_{32}	0.46	0.307	0.33	0.350	0.390

The cortical bone data of Rho (1992) are implemented in the FE models of this study. The data of Rho (1992) were chosen because he presented the most extensive set of data in comparison with other available studies. Rho (1992) tested a relatively large number of samples (8 femora and tibiae each) and reported the 9 elastic constants for each of these samples. These samples were taken from circumferential regions around the bones (i.e. at the anterior, posterior, medial, and lateral quadrants) as well as along the length of the bone (i.e at 30%, 50%, and 70% of full length). The stiffness data of the tibial and femoral cortex can be seen in Tables 4.6 and 4.7, respectively, and are well within the reported range of values found in other studies (Table 4.1).

A large range of cortical stiffness values has been reported by various investigators (Currey, 1970). One of the reasons for the large differences may be the embalming and drying of the cortical bone. Reilly and Burstein (1974) found that embalming of the bone changes the mechanical characteristics. The bone must also be kept moist as drying of the bone can induce cracks which changes the stress response and strength characteristics. Several studies (Reilly and Burstein, 1974; Reilly et al., 1974;

Carter and Hayes, 1977a) have taken this into account and their results contain a range of axial stiffness values from 14.1 to 22.1 GPa. The corresponding Young's modulus reported in the transversely isotropic (Reilly and Burstein, 1975) and orthotropic (Van Buskirk and Ashman, 1981; Ashman et al., 1984; Rho, 1992, 1996) studies all fall within this range.

Reilly and Burstein (1975) hypothesized that different areas of the bone had slightly different properties. Knets and Malmeisters (1975) performed tests on six different areas around the mid-diaphysis of a tibia and Van Buskirk and Ashman (1981) tested four different areas around the femoral shaft. Although no statistical information was given regarding these data, it appears that the data vary about 10 to 20% in differing regions of the cortical bone. Rho (1992, 1996) examined tibial cortical specimens ultrasonically along the length of the tibia (at 30%, 50%, and 70% of the length), and around the circumference (at anterior, lateral, medial, and posterior quadrants). He found no significant variation in elastic properties along the length and around the circumference of the cortical bone.

Several studies have investigated age-related variations in the mechanical properties of human cortical bone (Currey, 1979; Burstein et al., 1976; Currey and Butler, 1975; McCalden et al., 1993). Currey (1979) showed that the elastic modulus and bending strength both increased until about 30 years and declined thereafter. However, the mechanical properties of femoral and tibial tissues were found to differ in their response to the aging process after 30 years (Burstein et al., 1976). The data presented in the age studies above can be used to study the effects of aging bone in relation to knee reconstructions and total knee replacements using the tool reported in this study. The use of age-stiffness empirical relationships in the FE models is elaborated upon in Chapter 8.

Cortical bone is generally accepted as having viscoelastic properties. However, for most studies of stress and strain, it is reasonable to model cortical bone as linearly elastic (Van Buskirk and Ashman, 1981; Rho, 1996). Therefore, in this study, cortical bone is assumed to be linearly elastic.

4.1.2 Cancellous Bone

Cancellous bone is a cellular material made up of a connected network of rods and plates. The density of bone in a particular location depends on the magnitude of the applied loads, which can be explained by Wolff's Law, which states that bone remodels in response to the mechanical stresses it experiences so as to produce an anatomical structure best able to resist the applied stress. At low relative densities, the cells form an open network of rods (Fig. 4.3a). As the relative density increases, more material accumulates in the cell wall and the structure transforms into a more closed network of plates (Fig. 4.3b). It can be seen in Fig. 4.3c that in regions of higher stress, such as the femoral condyles, the cells form closed cell, plate-like structures.

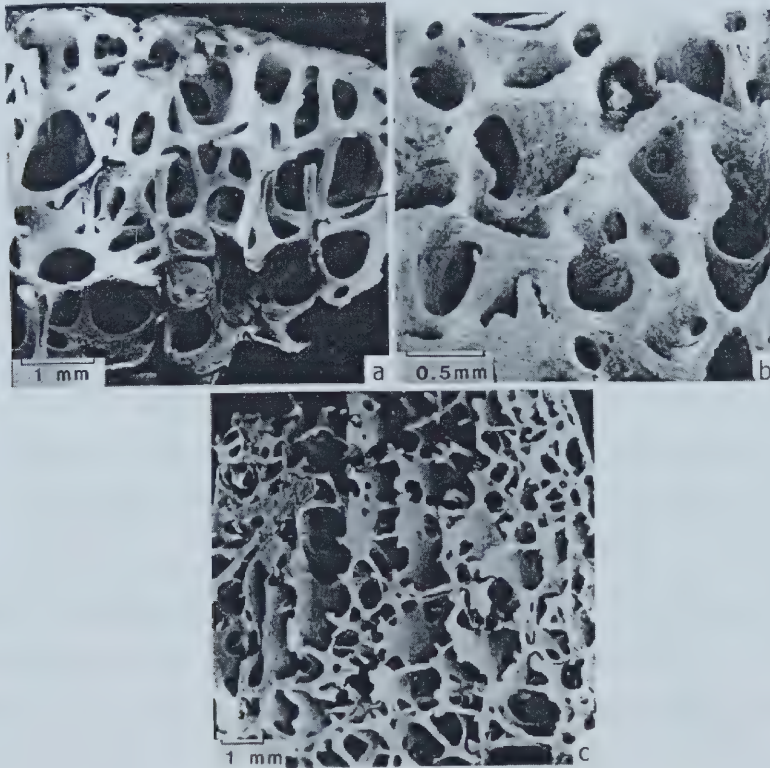


Figure 4.3: (a) Scanning electron micrograph of low density cancellous bone with an asymmetric rod-like structure. Specimen taken from the femoral head; (b) scanning electron micrograph of high density cancellous bone with an asymmetric plate-like structure. Specimen taken from the femoral head; (c) scanning electron micrograph of plate-like cancellous bone with columnar structure. Specimens taken from the femoral condyle. Reproduced with permission from Gibson (1985).

The symmetry of the structure in cancellous bone depends upon the direction of the applied loads. In bones where the loading is largely uniaxial, like the femur and tibia, the trabeculae often develop a columnar structure with cylindrical symmetry. The columns of bone are oriented in the vertical direction giving relatively high stiffness and strength in the direction of load with lower stiffness and strength in the transverse directions.

The nature of cancellous bone can be described by both material and structural properties. Structural properties are defined at the macroscopic level and contain the extrinsic properties of both trabeculae and pores, whereas material properties are defined at the microstructural level, containing the intrinsic properties of the trabecular struts only (Rho, 1992). The size of the typical trabeculae is small (approximately 0.3 mm diameter and 2 mm length) and specimens must span at least five intertrabecular lengths in order to relate local strains in trabecular bone to those on the apparent level (Harrigan et al., 1998). As the scope of this thesis focuses on the bone as a whole, this work is mainly interested in the macrostructure of cancellous bone rather than the microstructure. Therefore, the following discussion emphasizes the properties of cancellous bone rather than trabecular bone.

4.1.2.1 Elastic Modulus

The odd shape and small size of cancellous specimens make conventional testing procedures difficult, resulting in a variety of testing methods. Mechanical (i.e. tension, compression, and bending), ultrasonic, as well as optical techniques have been used to measure the mechanical properties. Values obtained using ultrasonic techniques show good correlation with those obtained by mechanical testing (Ashman et al., 1987). Table 4.2 summarizes the moduli of elasticity and Poisson's ratio of cancellous bone found in various studies.

Table 4.2: Summary of cancellous bone Young's modulus values found in literature. AP refers to the anterior-posterior direction; ML, the medial-lateral direction; and SI, the superior-inferior direction. Standard deviations are in parentheses.

Study	Direction	Young's Modulus (GPa)
FEMUR		
<i>Ducheyne et al. (1977)</i>	SI	0.059 – 2.942
<i>Rohlmann et al. (1980)</i>	SI	0.32
<i>Ciarelli et al. (1986)</i>	AP	0.051 – 0.428
	ML	0.060 – 0.388
	SI	0.150 – 0.387
<i>Ashman and Rho (1988)</i>	SI	0.96 – 2.17
<i>Rho (1992)</i>	AP	0.760 (0.589)
	ML	0.707 (0.555)
	SI	1.698 (0.993)
TIBIA		
<i>Hvid et al. (1983)</i>	SI	0.014 – 0.345
<i>Odgaard and Linde (1991)</i>	SI	0.69, 0.87
<i>Ashman et al. (1989)</i>	AP	0.35 (0.22)
	ML	0.46 (0.28)
	SI	1.11 (0.63)
<i>Williams and Lewis (1984)</i>	AP/ML	0.075, 0.066
	SI	0.28, 0.37
<i>Rho (1992)</i>	AP	0.202 (0.154)
	ML	0.232 (0.180)
	SI	0.769 (0.534)

There is considerable variation in the mechanical properties of cancellous bone due to characteristics such as porosity (Hodgkinson and Currey, 1992; Rho, 1992), mineral content (Van der Linden et al., 2001), trabecular orientation (Galante et al., 1970), and trabecular structure (Gibson, 1985). The resulting variation can be seen in the large range of reported Young's moduli of cancellous bone in both the femur and tibia (Table 4.2). Despite the large variations, there exist common material characteristics among cancellous bone.

Cancellous bone is known to exhibit anisotropic behaviour. Earlier studies reported cancellous bone as having transversely isotropic properties (Williams and Lewis, 1982; Ciarelli et al., 1986), while more recent studies have characterized cancellous bone as being orthotropic (Ashman et al., 1989; Rho, 1992, 1996). It is now generally accepted that cancellous bone exhibits orthotropic behaviour (Wirtz et al., 2003). One therefore must model the bone as anisotropic to have a realistic

representation of the bone stress states. Hence, previous 3D FE models incorporating isotropic bone (Lewis et al., 1982; Little et al., 1986; Hashemi and Shirazi-Adl, 2000) may not be adequate and therefore we have incorporated the available orthotropic material property data into our FE models to improve upon the state of the art.

It still remains very difficult to determine an appropriate range of values with which to compare Young's moduli of cancellous bone. The areas from which specimens are taken are not always reported and, due to the spatial variation in bones, comparisons will reveal large differences. The spatial variation of stiffness within a tibia can be as large as 800% (Rho, 1992); even within similar regions there are large differences. A comparison of the mapped tibial stiffnesses (Goldstein et al., 1983; Ashman et al, 1989; Rho, 1992) revealed a variation in the magnitude of the Young's modulus by as much as 100% in matching areas. A comparison of femoral maps produced by Rho (1992) and Ciarelli et al. (1986) revealed up to a tenfold difference in stiffness values. However, the pattern of the bone stiffness is similar in both tibial and femoral maps. For example, in the femur, posterior bone was stiffer than anterior bone, and lateral bone slightly stiffer than medial bone.

Bone is clearly stiffer beneath the centers of the load bearing areas of the condyles because of the higher stress concentrations (Rho, 1992). Within the distal femur, medial cancellous bone is significantly stiffer than lateral cancellous bone, and posterior bone is significantly stiffer than anterior bone (Rho, 1992). Bone in the condyles is also less stiff with increasing distance from the articular surface (Rohlmann et al., 1980; Ducheyne et al., 1977; Rho, 1992). Patterns of decreasing stiffness were found moving from epiphyseal bone to diaphyseal bone (Rho, 1992).

In the proximal tibia, it has been found that the medial stiffness is significantly greater than lateral stiffness (Rho, 1992, 1996). No significant difference was found between the stiffness of anterior and posterior bone (Rho, 1992). There is a general trend of decreasing bone stiffness as distance from the joint line increases. The bone just beneath the joint surface was found to be significantly stiffer than the bone 12mm distal to it, but successively distal layers were not found to be any stiffer (Rho, 1996).

To provide the best representation of bone properties in our FE model, we needed to consider cancellous bone as both orthotropic and highly inhomogeneous. Only one

study has published mapped orthotropic femoral stiffness data (Rho, 1992). Naturally, we chose to incorporate these data into our femur FE model. The same study (Rho, 1992) also provided mapped orthotropic tibial stiffness data. While Ashman et al. (1989) have also published similar data, for the sake of consistency with the femoral data, the tibial data of Rho (1992) were incorporated into our tibia FE model. Rho (1992) reported the stiffness of three transverse layers (10 mm apart) for femoral and tibial cancellous bone. 8 X 8 mm cancellous blocks were tested at various locations around each transverse layer. The bone divisions in the FE models of this study reflect as closely as possible the size and origins of the test samples. While Rho (1992) tested only three transverse layers of bone, our FE models required four layers of bone. Therefore, it was assumed that the material properties of the fourth transverse layer of bone were an average of the properties of the third layer.

4.1.2.2 Shear Modulus

To model the anisotropy of bone, shear modulus values are needed. There are no mapped shear modulus data, and published data regarding the shear modulus of the proximal tibia are scarce. Only Williams and Lewis (1982), Ashman et al. (1989), and Sonstegard and Matthews (1977) have reported the shear modulus of cancellous bone. The values reported are very similar. To the author’s best knowledge, there are no published data regarding the shear moduli of the distal femur. Table 4.3 summarizes the reported shear modulus values.

Table 4.3: Summary of shear modulus values found in literature. All reported values were for tibial specimens. Direction 1 is the anterior-posterior direction; direction 2 is the medial-lateral direction; direction 3 is the superior-inferior direction.

Study	Direction	Shear Modulus (GPa)
<i>Ashman et al. (1989)</i>	G ₁₂	0.098 (0.066)
	G ₁₃	0.13 (0.08)
	G ₂₃	0.17 (0.09)
<i>Williams and Lewis (1982)</i>	G ₁₃	0.13, 0.12
<i>Sonstegard and Matthews (1977)</i>	G ₂₃	0.15

Our FE models required the incorporation of spatial variation and, since there are no mapped shear modulus data, a method to represent the variation of shear modulus was necessary. Relationships between shear modulus and apparent density of cancellous bone have been reported by Ashman et al. (1989). The equations can be seen in the section 4.2.2. Therefore, rather than using specific shear modulus values reported in literature, the shear modulus-apparent density relationships of Ashman et al. (1989) were used. The resulting shear moduli incorporated into our models can be seen in Tables 4.6 and 4.7; they fall within the range of reported shear moduli.

4.1.2.3 Poisson's Ratio

In addition to Young's and shear moduli, Poisson's ratios are needed to characterize the anisotropy of bone. However, very scarce Poisson's ratio experimental data are available. This is due to the fact that Poisson's ratio of cancellous bone is very difficult to measure experimentally. McElhaney et al. (1970) studied the Poisson's ratio of cancellous bone using a porous block model. They reported a Poisson's ratio of 0.3 in the cancellous bone of the femur. Williams and Lewis (1982) used a combined experimental-numerical technique to estimate the Poisson's ratio of trabecular bone in the proximal 3 cm of the tibia. Although cancellous bone is known to be highly anisotropic, no experimental data has been published regarding the orthotropy of the Poisson's ratio of cancellous bone (Kowalczyk, 2003). Table 4.4 summarizes the literature reporting Poisson's ratio values.

Table 4.4: Summary of Poisson's ratio values found in literature. All reported values were for tibial specimens. Direction 1 is the anterior-posterior direction; direction 2 is the medial-lateral direction; direction 3 is the superior-inferior direction.

Study	Direction	Poisson's Ratio
<i>Williams and Lewis (1982)</i>	ν_{12}	0.52, 0.51
	ν_{31}	-0.07, -0.06
<i>McElhaney et al. (1970)</i>	ν_3	0.3

While the reported values are quite different between the two studies, because our FE models considered anisotropy, the data of Williams and Lewis (1982) were incorporated into our models. The Poisson's ratios were assumed to be the same in both the femur and tibia FE models.

As with cortical bone, cancellous bone also exhibits viscoelastic properties. However, in studies where bone deformation is small, cancellous bone can be considered linearly elastic with reasonable accuracy (Rho, 1996; Cowin, 1989). In this study, cancellous bone is assumed to be linearly elastic.

4.1.3 Subchondral Bone

The subchondral plate is a very thin layer of bone separating the articular cartilage from the trabecular (or cancellous) bone of the metaphysis (Radin and Rose, 1986). It is composed of two layers: a mineralized layer (contiguous with the articular cartilage), and a layer of lamellar bone (connected with the marrow). The tidemark demarcates the interface between the subchondral plate and the articular cartilage. In the knee, the subchondral plate is thickest at the tibial plateau because of its weight bearing function. The function of the subchondral plate is to provide support for the overlying articular cartilage (Duncan et al., 1987). Hayes et al. (1978) have also observed that the subchondral bone transfers loading to the cortical bone.

The published data reporting the mechanical properties of subchondral bone are rare because of the difficulty of subjecting subchondral bone to conventional mechanical testing. Choi et al. (1990) tested subchondral microspecimens from the medial side of the proximal tibia. They reported a stiffness of 1.15 GPa. Low stiffness values have also been reported by other investigators. Murray et al. (1984) found a mean stiffness of 3.06 GPa, and Brown and Vrahas (testing subchondral bone from the femoral head) reported a stiffness of 1.37 GPa. To the author's knowledge, no literature has been published regarding the anisotropy of subchondral bone. Table 4.5 summarizes the literature which reported the Young's modulus of subchondral bone. Only Brown and Vrahas (1984) are known to have reported the Poisson's ratio of subchondral bone. They have estimated it

to be between 0.2 and 0.4, and a value of 0.25 was subsequently assumed for subchondral bone in a FEM study by Brown and DiGioia (1984).

While the anisotropy of bone properties has been emphasized in this study, the lack of experimental data regarding the material properties of subchondral bone does not allow for its full characterization. As such, subchondral bone must be assumed to be isotropic and homogeneous. Other FE studies (Hayes et al., 1978) have also assumed subchondral bone as isotropic and homogeneous. The reported values for the subchondral bone of the femur (specifically, the data of Choi et al. (1990)) and the tibia (Brown and Vrahas, 1984) are very similar. With Brown and Vrahas (1984) reporting the only experimental Poisson's ratio, naturally, we used their data. Our FE model assumes the subchondral bone as having a stiffness of 1.15 GPa and a Poisson's ratio of 0.25.

Table 4.5: Summary of reported stiffness values of subchondral bone. ML refers to the medial-lateral direction; and SI, the superior-inferior direction.

Study	Direction	Young's Modulus (GPa)
<i>TIBIA</i>		
<i>Choi et al. (1990)</i>	ML	1.15
<i>Murray et al. (1984)</i>	SI	3.06
<i>FEMUR</i>		
<i>Brown and Vrahas (1984)</i>	SI	1.37

4.2 Implementation of Material Properties

4.2.1 Cortical Bone

The cortical bone in the femur and tibia models was assumed to be orthotropic and inhomogeneous. To represent the inhomogeneity of the cortical bone, the cortical bone of the each FE model was divided into four quadrants (anterior, posterior, medial, and lateral). The four quadrants represent as closely as possible the mapped cortical bone data presented by Rho (1992). In ANSYS®, this was performed by using Boolean operations at four arbitrarily selected locations around the circumference of the cortical bone to divide the cortex into anterior, posterior, medial, and lateral quadrants. Upon division, the adjacent volumes were assumed to have continuity of displacement. This

was verified in Chapter 7. Figure 4.4 shows the divisions of the tibia FE model. Volume numbers 2, 22, 24, and 25 represent the cortical bone. The femur FE model was divided in a similar fashion to the tibia FE model (Fig 4.5). Volumes 3, 23, 24, and 25 represent the cortical bone. As the properties of cortical bone are more homogeneous along the length than around the periphery (Rho, 1992), only one transverse layer of cortical bone was modeled for both femur and tibia. To represent orthotropy, nine elastic constants for each quadrant were assigned to both the femur and the tibia based on experimental values found by Rho (1992). The cortical bone values assigned to the tibia and femur FE models are presented in Tables 4.6 and 4.7, respectively.

4.2.2 Cancellous Bone

The heterogeneity of the cancellous bone was represented in the FE models by dividing the cancellous bone into different sections. Different volumes were created using Boolean operations identical to those used in the cortical bone. Divisions were made based on the maps provided by Rho (1992). Transverse layers were sectioned as closely as possible at 10 mm intervals; within each layer, 8 X 8 mm sections were created whenever possible.

As stated above in Section 4.1.2, the cancellous bone in the femur and tibia was assumed to be orthotropic and inhomogeneous. The elastic moduli assigned to the models were obtained from experimental data published by Rho (1992). These data included the orthotropic modulus of elasticity data for each sample of cancellous bone.

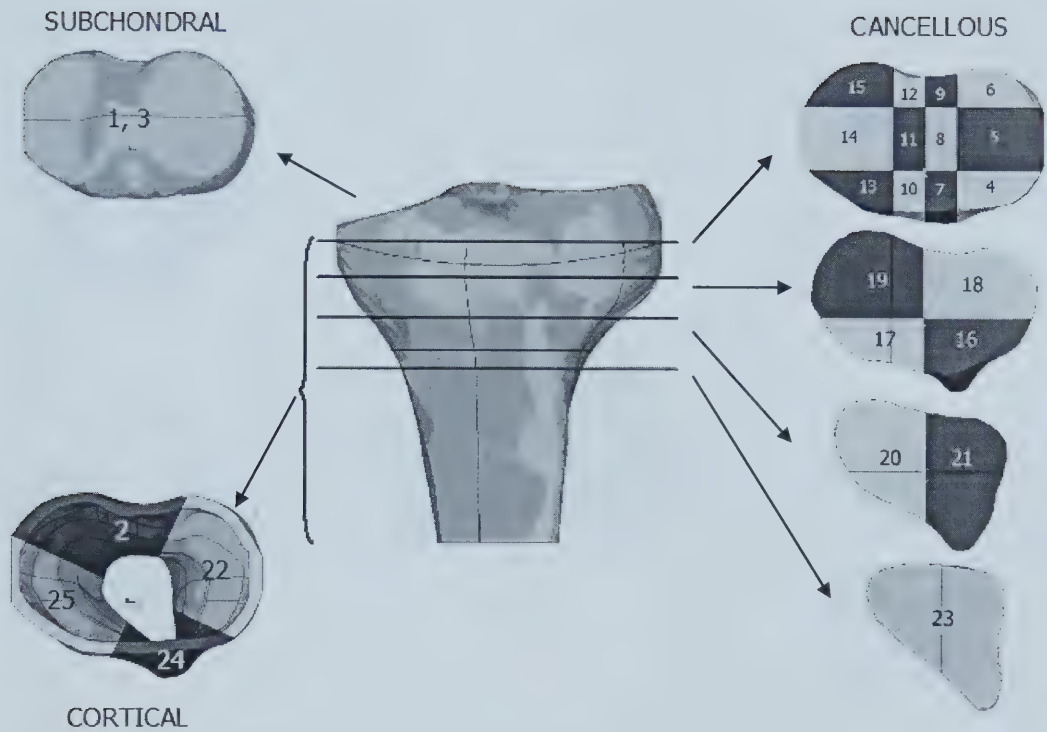


Figure 4.4: The bone divisions within the tibia FE model. There are four transverse sections of cancellous bone, four quadrants of cortical bone, and one section of subchondral bone. The volume numbers were assigned their corresponding values in Table 4.6. The section of subchondral bone contains two volumes (1 and 3) with volume 1 enclosing volume 3.

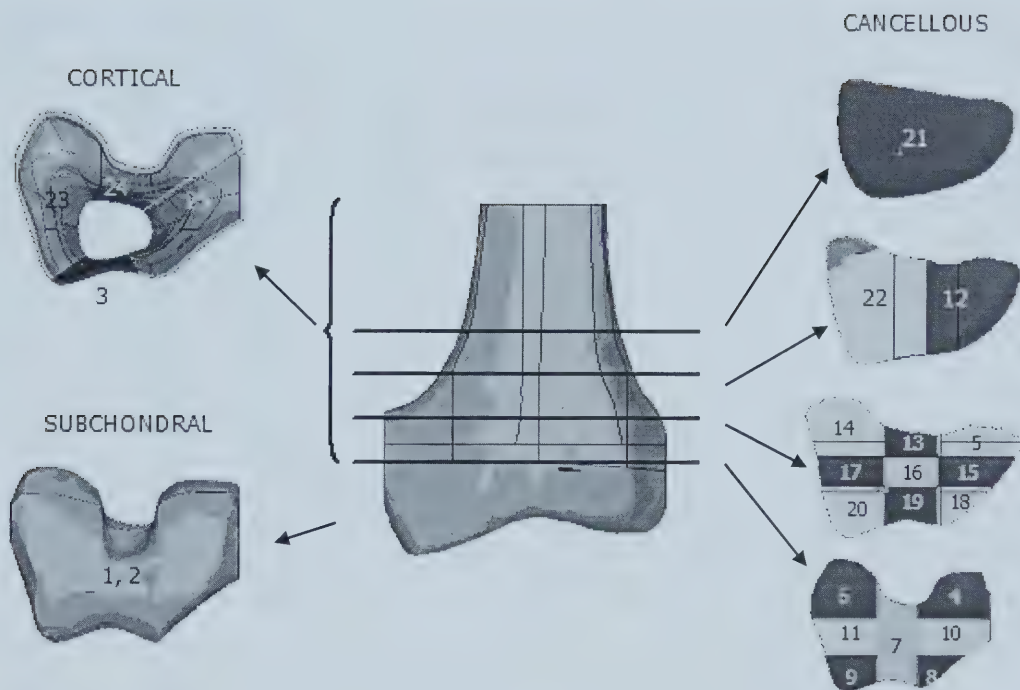


Figure 4.5: The bone divisions within the femur FE model. There are four transverse sections of cancellous bone, four quadrants of cortical bone, and one section of subchondral bone. The volume numbers were assigned their corresponding values in Table 4.7. The section of subchondral bone contains two volumes (1 and 2) with volume 2 enclosing volume 1.

Table 4.6: Elastic constants assigned in the tibia FE model. E is Young's modulus; G is shear modulus; ν is Poisson's ratio. E and G are in units of MPa. Direction 1 is in the anterior-posterior direction; direction 2 is in the medial-lateral direction; direction 3 is in the superior-inferior direction. Volume numbers refer to the bone sections in Fig 4.4. Young's moduli values were average bone properties reported by Rho (1992); shear moduli were calculated as described in the text; Poisson's ratios were reported by Williams and Lewis (1982).

Volume	E_1	E_2	E_3	G_{12}	G_{13}	G_{23}	ν_{12}	ν_{13}	ν_{23}
1	N/A	N/A	1150	N/A	N/A	N/A	0.25	N/A	N/A
2	12500	12200	19400	4400	5400	5900	0.432	0.236	0.239
3	N/A	N/A	1150	N/A	N/A	N/A	0.25	N/A	N/A
4	440	492	1759	125	213	181	0.52	-0.02	-0.07
5	294	293	1132	75	134	113	0.52	-0.02	-0.07
6	350	403	1226	93	162	137	0.52	-0.02	-0.07
7	283	329	934	71	128	109	0.52	-0.02	-0.07
8	121	108	362	24	48	40	0.52	-0.02	-0.07
9	196	209	734	48	90	76	0.52	-0.02	-0.07
10	202	252	731	52	97	82	0.52	-0.02	-0.07
11	122	131	389	26	53	44	0.52	-0.02	-0.07
12	200	248	927	58	106	89	0.52	-0.02	-0.07
13	283	376	1006	78	139	118	0.52	-0.02	-0.07
14	171	178	722	43	82	69	0.52	-0.02	-0.07
15	322	437	1302	96	168	142	0.52	-0.02	-0.07
16	100	113	433	25	49	41	0.52	-0.02	-0.07
17	103	114	452	25	51	42	0.52	-0.02	-0.07
18	110	117	497	27	51	45	0.52	-0.02	-0.07
19	94	103	415	23	46	38	0.52	-0.02	-0.07
20	75	140	389	24	48	40	0.52	-0.02	-0.07
21	97	140	433	27	53	44	0.52	-0.02	-0.07
22	12000	12500	21900	4100	5600	5600	0.419	0.220	0.213
23	86	140	411	25	51	42	0.52	-0.02	-0.07
24	11800	14100	19100	4200	5500	6300	0.306	0.286	0.262
25	11400	11700	19600	3900	5300	5600	0.437	0.228	0.223

Table 4.7: Elastic constants assigned in the femur FE model. E is Young's modulus; G is shear modulus; ν is Poisson's ratio. E and G are in units of MPa. Direction 1 is in the anterior-posterior direction; direction 2 is in the medial-lateral direction; direction 3 is in the superior-inferior direction. Volume numbers refer to the bone sections in Fig 4.5. Young's moduli values were average bone properties reported by Rho (1992); shear moduli were calculated as described in the text; Poisson's ratios were reported by Williams and Lewis (1982).

Volume	E_1	E_2	E_3	G_{12}	G_{13}	G_{23}	ν_{12}	ν_{13}	ν_{23}
1	N/A	N/A	1150	N/A	N/A	N/A	0.25	N/A	N/A
2	N/A	N/A	1150	N/A	N/A	N/A	0.25	N/A	N/A
3	11200	11600	19600	3800	5200	4900	0.457	0.218	0.213
4	1654	1173	2676	326	434	501	0.52	-0.02	-0.07
5	599	669	1516	147	210	245	0.52	-0.02	-0.07
6	1828	1842	3098	428	557	640	0.52	-0.02	-0.07
7	1071	1054	1941	238	326	412	0.52	-0.02	-0.07
8	606	604	1627	145	207	242	0.52	-0.02	-0.07
9	1123	1096	2468	267	361	418	0.52	-0.02	-0.07
10	1093	1129	2328	262	356	412	0.52	-0.02	-0.07
11	1118	1076	2110	251	341	396	0.52	-0.02	-0.07
12	190	171	678	43	68	81	0.52	-0.02	-0.07
13	312	386	810	75	113	134	0.52	-0.02	-0.07
14	723	762	1991	183	256	298	0.52	-0.02	-0.07
15	264	280	967	66	102	120	0.52	-0.02	-0.07
16	224	183	683	46	73	87	0.52	-0.02	-0.07
17	599	437	1487	123	178	209	0.52	-0.02	-0.07
18	420	416	1241	100	147	173	0.52	-0.02	-0.07
19	571	527	1409	127	184	216	0.52	-0.02	-0.07
20	917	643	2320	198	275	320	0.52	-0.02	-0.07
21	277	228	945	63	97	115	0.52	-0.02	-0.07
22	363	285	1212	82	123	145	0.52	-0.02	-0.07
23	12200	12600	21300	4200	5200	5700	0.424	0.249	0.245
24	11500	12500	21300	3900	5500	4800	0.424	0.221	0.208
25	12400	13100	19600	4400	5000	5500	0.403	0.252	0.245

To represent the inhomogeneity of the cancellous bone along the length of the tibia, the FE model was divided into four transverse layers (Fig 4.4). The elastic moduli of the proximal three layers of cancellous bone were modeled on the mapped experimental data (Rho, 1992). As experimental data from Rho (1992) were only available for the proximal three layers, material properties of the fourth layer of cancellous bone were assumed to be an average of the properties of the third layer. To represent the spatial variation peripherally, each of the three transverse layers was divided into numerous volumes. The sections were divided in a similar manner to the mapped data (Rho, 1992). Each section of cancellous bone was assigned a different set of orthotropic material properties. The elastic constants assigned in the tibia FE model are shown in Table 4.6, and the corresponding Fig 4.4. Volumes 4 to 21 represent the cancellous bone in the tibia.

The spatial variation of femoral cancellous bone was represented in a similar manner as that in the tibia FE model. The different placement of the volumes between the femur and tibia FE models are representative of the different maps for the femur and tibia experimental data. Similar to the tibia, material properties for the fourth transverse layer were assumed to be an average of the properties of the third layer. The elastic constants assigned in the femur FE model are shown in Table 4.7, and the corresponding Fig 4.5. Volumes 4 to 22 represent the cancellous bone in the femur model.

The shear moduli for each cancellous bone volume were calculated from empirical relationships reported by Ashman et al. (1989). Three empirical equations related Young's modulus to apparent density for three mutually perpendicular directions (Eqs 4.1 to 4.3); three other equations related shear modulus to apparent density for the same directions (Eqs 4.4 to 4.6).

$$E_1 = 0.52\rho^{1.16} \quad (4.1)$$

$$E_2 = 0.79\rho^{1.14} \quad (4.2)$$

$$E_3 = 2.84\rho^{1.07} \quad (4.3)$$

$$G_{12} = 0.08\rho^{1.26} \quad (4.4)$$

$$G_{13} = 0.22\rho^{1.15} \quad (4.5)$$

$$G_{23} = 0.29\rho^{1.13} \quad (4.6)$$

where E is the Young's modulus (MPa), G is the shear modulus (MPa), and ρ is the apparent density (kg/m^3); directions 1, 2, and 3 are the anterior-posterior, medial-lateral, and superior-inferior directions, respectively.

For each volume of bone, an apparent density was calculated using the Young's moduli – apparent density relationships. Using this apparent density, the shear moduli (for three directions) were calculated using the shear moduli – apparent density relationships.

The same empirical relationships were also used to determine the shear moduli for femoral cancellous bone. Although distal femoral cancellous bone was found to be more than twice as stiff (a significant difference) as proximal tibial cancellous bone (Rho, 1992), due to the lack of published data regarding the shear moduli, the values found using the tibial empirical relationships were felt to be the most representative of femoral cancellous bone as they are from the same long bone family.

Identical Poisson's ratios were used in femur and tibia FE models. The transversely isotropic ratios reported by Williams and Lewis (1982) were assigned to the models. These values are very similar to those implemented by Askew and Lewis (1981) in their 2D FE tibia model.

4.2.3 Subchondral Bone

One layer of subchondral bone was modeled in each of the femur and tibia FE models. In the tibia FE model, the subchondral bone is represented by volumes 1 and 3; in the femur FE model, it is represented by volumes 1 and 2. Because of the original surface geometry, two volumes were created in each layer. However, the two volumes share identical material properties and, therefore, should be considered a single entity. As discussed above in Section 4.1.3, there are few data regarding the mechanical properties of subchondral bone. As there are no data reporting the anisotropy of subchondral bone, its mechanical properties were assumed isotropic. In addition, subchondral bone was assumed homogeneous and linearly elastic. The same set of data

was used for both the femur and the tibia. The Young's modulus of the subchondral plate was assumed to be 1.15 GPa (Choi et al., 1990) and the Poisson's ratio was assumed to be 0.25 (Brown and DiGioia, 1984).

Chapter 5: Meshing of Finite Element Model

5.1 Selection of Element Type and Size

Meshing three-dimensional finite element models of long bones can be accomplished either through manual or automatic mesh generation. The manual generation of meshes is a complex and time-consuming task. For this reason, mesh refinements or convergence checks are rarely reported due to the large amount of time needed to produce a single mesh (Viceconti et al., 1998). To reduce the time to generate meshes, programs (automeshers) have been developed that can automatically generate FE meshes of an object, starting from its geometric description (Viceconti et al., 1998). Automeshers are now commonly available in commercial FE pre-processors. The commercial finite element analysis package ANSYS® is capable of automatically generating meshes using tetrahedral and brick elements, among other types of elements.

Various element types have been used to mesh three-dimensional solid models of the femur and tibia. These various elements include eight-node parallelepiped (Askew and Lewis, 1982), eight-node (hexahedral) brick (Nambu et al., 1991; Keyak et al., 1990, 1992, 1993; Hashemi and Shirazi-Adl, 2000; Rohlmann et al., 1982), six-node prism (Hashemi and Shirazi-Adl, 2000), and four- and ten-node tetrahedral elements (Polgar et al., 2001). Hexahedral meshes are more commonly used but generally require more computational power than meshing with tetrahedral elements. In the meshing of irregular geometries such as the femur and tibia, tetrahedral elements are better suited than hexahedral elements (Viceconti et al., 1998). In comparing four- and ten-node tetrahedral elements, Polgar et al. (2001) stated that four-node linear tetrahedral elements should be avoided and quadratic tetrahedral elements are better suited for finite element analysis of the human femur.

An investigation was required to determine an appropriate element type to be used for meshing of the solid models. The ANSYS® pre-processor was used to mesh a solid tibia model with four different element types. Automatic mesh generation was used with eight- and twenty-node brick elements, and four- and ten-node tetrahedral elements. Convergence analyses were first performed to determine an appropriate element size for

comparison of the four element types. Convergence of stresses and displacements were evaluated in both the femur and tibia FE models.

Stress convergence was performed by increasing the number of degrees-of-freedom (DOF) in the FE model and observing the solution at critical points where the largest changes occur. A convergence test was performed on both the femur and the tibia FE models. The DOFs were increased by meshing the FE model with 5 different element sizes (5, 4, 3, 2, and 1.5 mm). A 5 mm mesh contained approximately 65,000 nodal DOFs whereas a 1.5 mm mesh contained approximately 1,700,000 nodal DOFs. Elements larger than 5 mm did not represent the curved surfaces of the femur and the tibia adequately, resulting in large element distortions and therefore unsatisfactory meshes. Loading was identical for all mesh sizes with a relevant physiological load of 343 N divided between two nodes located medially and laterally on the joint surface. The load of 343 N was obtained by assuming a 70 kg person standing with their weight divided over both legs.

The stress convergence test for the tibia FE model containing anisotropic and inhomogeneous material properties was performed. A bone tunnel was placed in the tibia to demonstrate the application of the FE model to anterior cruciate ligament reconstruction surgery. In this model, six locations were observed: 4 on the cortical bone surface (points A to D), and 2 in the cancellous bone (points E and F) (Fig 5.1). Point A is on the lateral cortex; B, the posterior cortex; C, the medial cortex; and D, the anterior cortex. Point E is in the proximity of the bone tunnel while point F is directly on the bottom edge of the tunnel. The locations were selected, firstly, because they were in the distal tibia and bottom edge of the tunnel where stresses are expected to be relatively high. Convergence is observed in critical areas; it would be safe to consider that convergence is also achieved in less critical areas. Secondly, to allow for the best comparison possible, the six nodes were selected because they had identical coordinates across all the mesh densities. The von Mises stresses at these six locations were compared for the five different meshed models. The stresses were plotted against the nodal degrees of freedom (Fig 5.2). The von Mises stresses in each of the six locations converged as the number of DOFs increased. It was concluded that a 2 mm element size

(containing 696,603 nodal degrees of freedom) was the most computationally efficient size without noticeable loss in accuracy compared to finer mesh sizes.

A stress convergence test was also performed on a femur FE model with a bone tunnel using a similar strategy as above. The femur model was monitored at eight locations: 4 on the cortical bone (points A to D) and 4 in cancellous bone (points E to H) (Fig 5.3). Point A is on the anterior cortex; B, the lateral cortex; C, the medial cortex; D, the posterior cortex. Points E and H are on the surface of the bone tunnel at mid-length and full-length of the tunnel, respectively. Points F and G are in the proximity of points E and H, respectively. The eight points were selected using the same strategy as described above for the tibia FE model. There was clear convergence for points A to G (Fig 5.4). Stresses at only a few points in the bottom edge of the tunnel (e.g. point H) showed oscillatory behaviour with increasing nodal degrees of freedom (perhaps for these few points, a local refinement of mesh would resolve this situation). Overall, the femur FE model converges with 2 mm elements (with 871,599 nodal degrees of freedom).

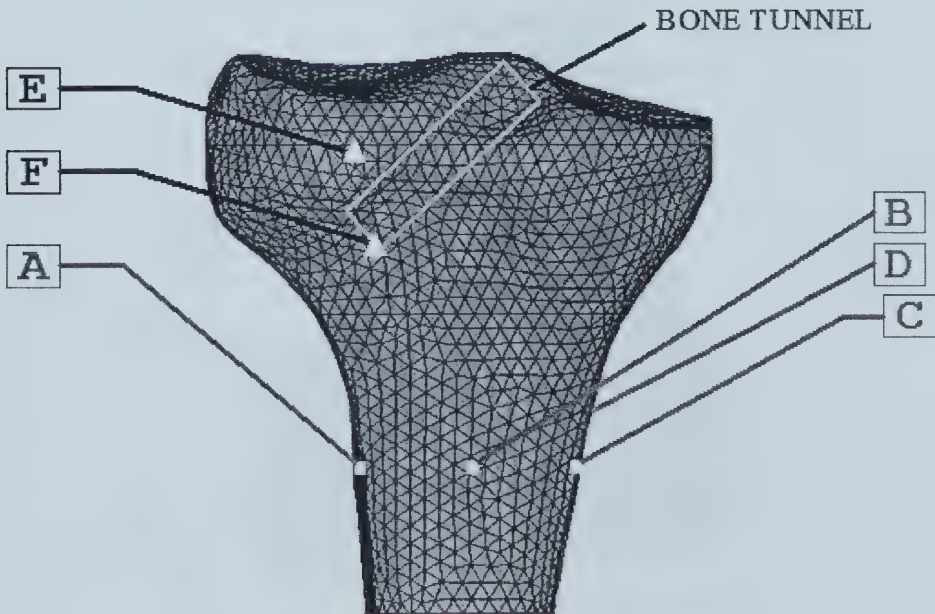


Figure 5.1: The six points used to monitor von Mises stress on the tibia FE model. Points A to D are on the surface of the cortex (point D is on the anterior cortex, directly opposite to point B); points E and F are in the cancellous bone. The rectangle represents the placement of the tunnel inside of the bone in the model.

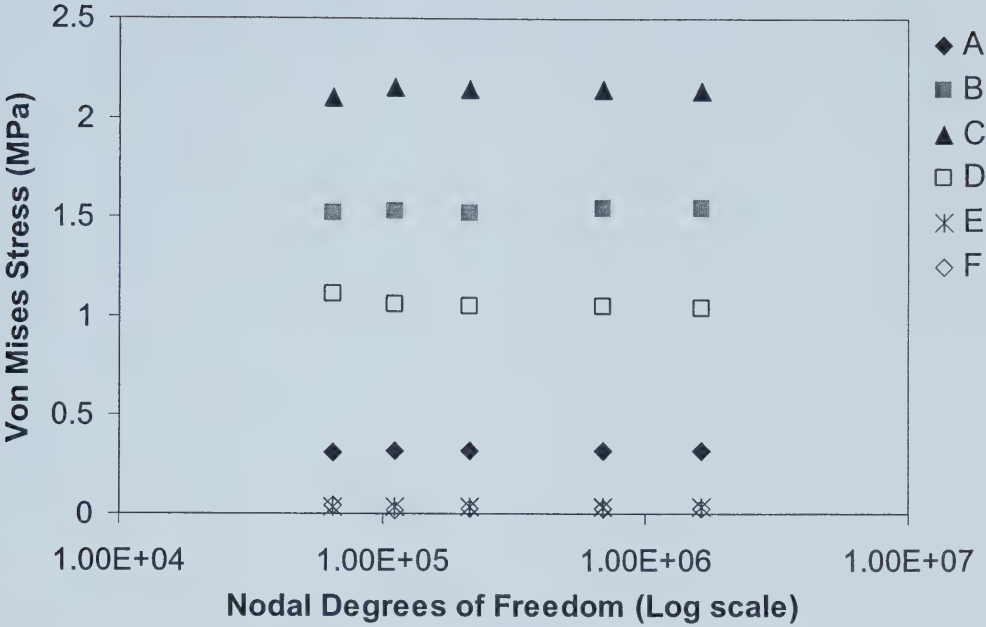


Figure 5.2: Convergence plot for the tibia FE model. Von Mises stresses for six locations (shown in Fig 5.1) are plotted against the nodal degrees of freedom.

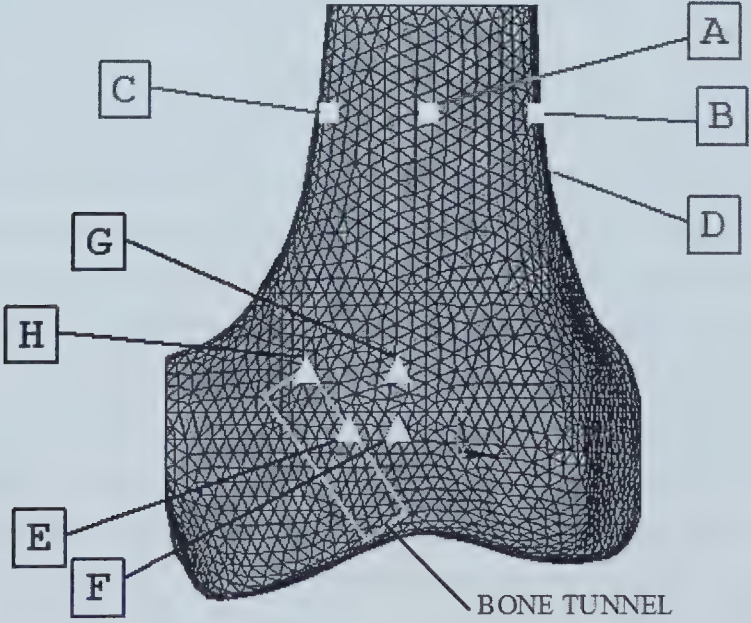


Figure 5.3: The eight points used to monitor von Mises stress on the femur FE model. Points A to D are on the surface of the cortex (point D is on the posterior cortex, directly opposite to point A); points E to H are in the cancellous bone. The rectangle represents the placement of the tunnel inside of the bone in the model.

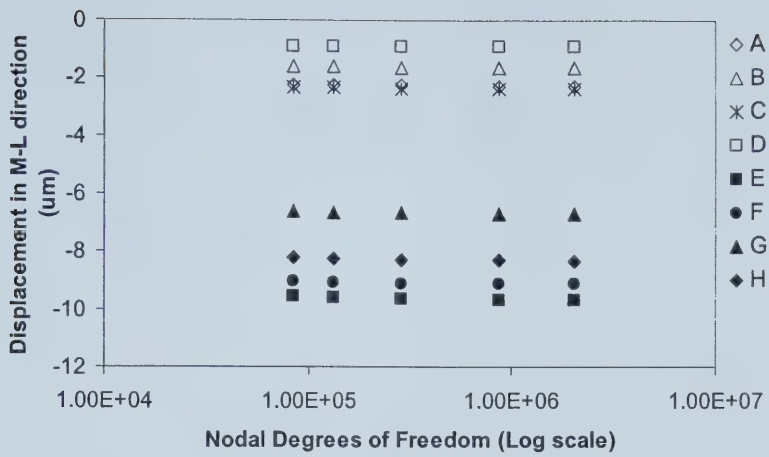


Figure 5.4: Convergence plot for the femur FE model. Von Mises stresses for eight locations (shown in Fig 5.3) are plotted against the nodal degrees of freedom.

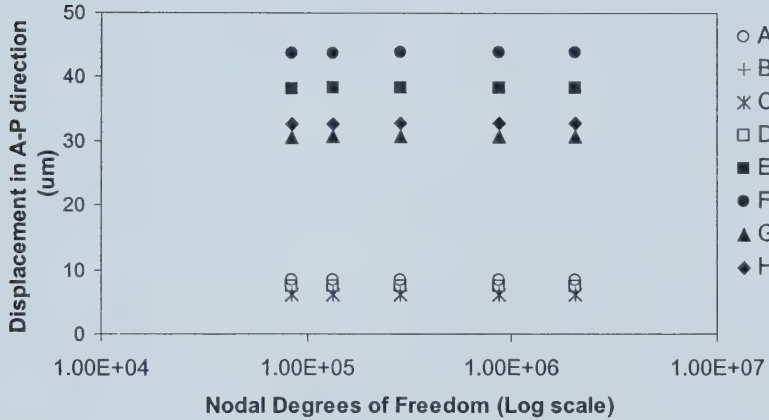
Similar to the stress convergence tests, a convergence test examining the displacement with increasing degrees of freedom was performed to ensure accuracy of the FE models. Identical loading and boundary conditions to the above stress convergence analysis were used. Displacements were monitored at the same locations as the stresses (i.e. Fig 5.3 for the femur FE model; Fig 5.1 for the tibia FE model). In the femur FE model, excellent convergence was found. There was little change for all points on the femur in the superior-inferior direction (Fig 5.5) with increasing nodal degrees of freedom. Similar results were found for the displacements in the medial-lateral and anterior-posterior direction. Displacement changes of less than 1% were found for all mesh densities at all eight locations. In the tibia FE model, good convergence was found. Differences in the anterior-posterior and superior-inferior direction were less than 2% for all mesh densities at all six locations (Fig 5.6). In the medial-lateral direction, the displacement at point B converged to a difference of 5% for meshes of 1.5 and 2 mm (Fig 5.6a). Differences in displacements in the other directions at point B were approximately

zero, verifying good convergence at this location. Displacement changes of less than 1% were found for points A, C, D, E, and F in the medial-lateral direction. The displacements of the femur were considered converged at an element size of 5 mm and, for the tibia, at 2 mm. The displacements were seen to converge more rapidly than the stresses. As calculations of stress are dependent on displacements, displacements must always converge before stresses can. The anisotropic properties of the bone result in more a complex set of stresses, which require a larger number of degrees of freedom to converge.

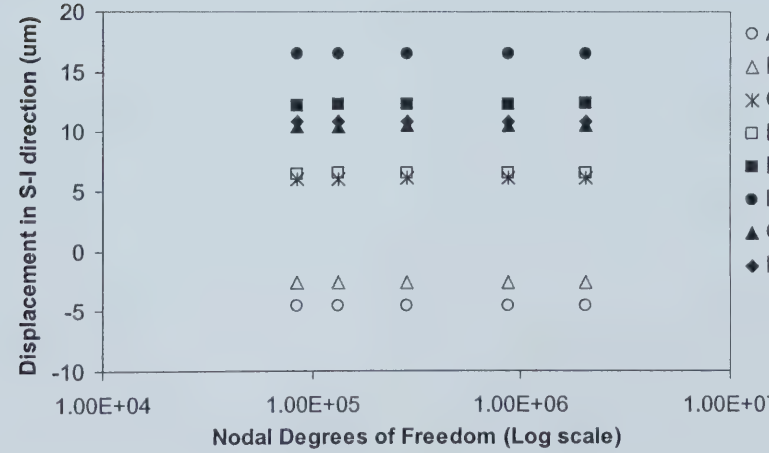
The above analysis shows that application of 2 mm tetrahedral elements would provide accurate solutions for the femur and tibia FE models. It was assumed that 2 mm elements for all four element types would also provide accurate results. A tibia FE model was alternately meshed with four- and ten-node tetrahedral elements, and eight- and twenty-node brick elements. Mesh generation was unsuccessful with both eight-node brick and four-node tetrahedral elements. In generating the mesh with four-node tetrahedral elements, there were no difficulties in generating the mesh for the cancellous and cortical bone volumes. However, the automeshing had difficulty meshing the subchondral surface. In particular, a mesh could not be generated to connect the cortical and subchondral surfaces. With the eight-node brick elements, meshes could not be generated for any of the volumes at the sizes considered here. Mapped meshing must be used to automatically generate a mesh with brick elements. Because of the irregular geometry of the tibia, mapped meshing cannot be generated. The difficulties in meshing with the linear elements may be due to the absence of mid-side nodes on these elements, making it very difficult to mesh the curved surfaces of the tibia. An attempt to mesh the volumes with smaller sizes of these elements (e.g. 1 mm elements) was also unsuccessful. The same difficulties as described above were encountered.



(a)

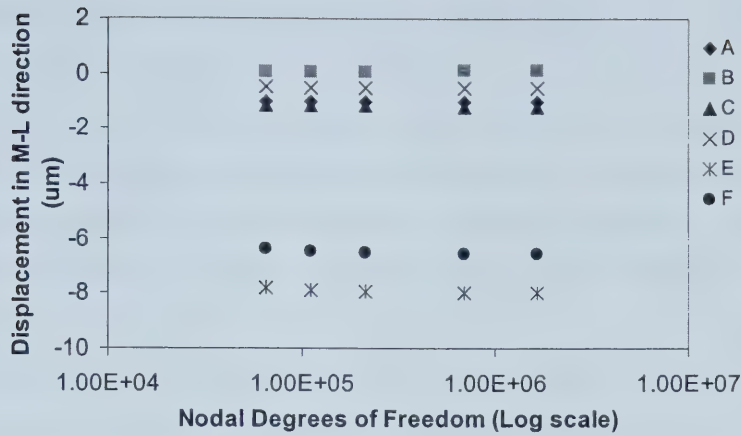


(b)



(c)

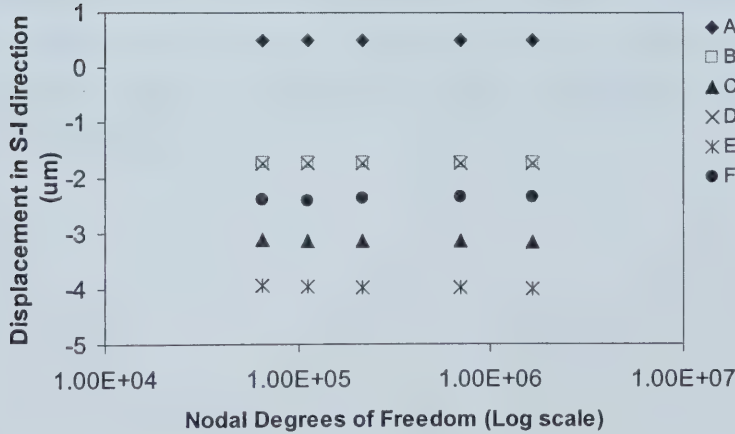
Figure 5.5: The displacement in the (a) medial-lateral direction, (b) anterior-posterior direction, and (c) superior-inferior direction for eight monitored points (see Fig 5.3) on the femur FE model. As the number of nodal degrees increase, the displacements converge.



(a)



(b)



(c)

Figure 5.6: The displacement in the (a) medial-lateral direction, (b) anterior-posterior direction, and (c) superior-inferior direction for six monitored points (see Fig 5.1) on the tibia FE model. As the number of nodal degrees increase, the displacements converge.

Meshing with ten-node tetrahedral and twenty-node brick elements produced successful meshes with minimal element shape distortions (0.01% of the elements were distorted). Only when a triangular surface mesh was used in conjunction with twenty-node brick elements could a successful mesh be generated. When a quadrilateral surface mesh was used, meshes for all the volumes could not be generated. This is because mapped meshing is necessary when using twenty-node brick elements with quadrilateral surface meshing. A comparison of the displacements and von Mises stresses from the ten-node tetrahedral and twenty-node brick meshes revealed identical results. An identical number of nodes was produced for the two sets of meshes. This may be explained by the degeneration of the twenty-node brick elements into ten-node tetrahedral elements, thus producing two identical meshes. The degeneration occurs by the brick element collapsing into a tetrahedral element; two of the quadrilateral faces disappear and, as a result, 11 nodes merge into 1 node. Therefore, the brick mesh will essentially have the same number of nodes as a tetrahedral mesh.

Viceconti et al. (1998) found that tetrahedral meshing yields the best results in cases where a solid model of bone is available. Subsequent FE models in this study were meshed with 2 mm tetrahedral elements except where more specific information is provided. An average meshed tibia model contains 168,407 elements of size 2mm, and 696,603 nodal degrees of freedom. The average meshed femur model contains 207,665 elements of size 2mm, and 871,599 nodal degrees of freedom. The number of elements produced in these meshes is substantially more than those used in other 3D FE models, e.g. the FE model of Hashemi and Shirazi-Adl (2000) was meshed with 10,900 elements and contained 14,300 nodes.

Chapter 6: Loading of Finite Element Model

6.1 Background

A variety of loading conditions has been implemented in FE models. These include point loading (Eibeck et al., 1979; Askew and Lewis, 1981; Lewis et al., 1982), Fourier series (Hayes et al., 1978; Van Campen et al., 1979; Shrivastava et al., 1982; Murase et al., 1983), uniform (Nambu et al., 1991; Hashemi and Shirazi-Adl, 2000) and non-uniform distributed loading (Little et al., 1986). To obtain accurate internal stress analyses of the bone, accurate and physiologically representative loading conditions must be assigned to FE models. Physiological loading within the tibiofemoral joint is not simple and cannot be adequately represented by point loading, Fourier series loading, or uniform distributed loading. Various aspects of loading must be represented for accurate loading of FE models: contact pattern, load distribution, contact area, and magnitude of loading.

Menisci have been shown to carry a significant portion of loading applied to the knee, and its removal greatly increases the pressure on the joint surface (Fukubayashi and Kurosawa, 1980). A more representative loading condition can be implemented by including menisci into the models, which can be considered for future models. As the current models use loading representative of meniscectomized knees, the discussion of contact areas and patterns below focuses on contact with the menisci removed.

The contact patterns of the tibiofemoral joint have been experimentally examined by Kettlekamp and Jacobs (1972), Walker and Hajek (1972), Fukubayashi and Kurosawa (1980), and Brown and Shaw (1984). Schematic diagrams of the contact patterns found in the study by Fukubayashi and Kurosawa (1980) showed the medial and lateral patterns as being elliptical when the menisci are removed. Contact pressures were greatest near the center of the ellipse with successively lower pressures surrounding it. The shape of the contact areas was also reported to be generally elliptical by Walker and Hajek (1972). Brown and Shaw (1984) noted that loading pattern on the femoral condyles involved a predominant central peak with rather smooth decay toward the contact periphery. Non-uniform distributed loading can be seen to occur physiologically. Therefore, loading of

FE models using point loads, Fourier series, or even uniform distributed loading would be considered inaccurate. Only Little et al. (1986) have used non-uniform distributed loading. However, the contact pattern adopted was trapezoidal in shape rather than elliptical as seen physiologically. The loading used in the FE models of this study used non-uniform distributed loading with an elliptically shaped pattern to better represent physiological loading. The loading approximated the contact area and corresponding pressure distribution patterns reported by Fukubayashi and Kurosawa (1980), who presented the most detailed information of the above studies.

The contact areas of the knee have been evaluated using numerous methods including casting (Fukubayashi and Kurosawa, 1980; Walker and Hajek, 1972), piezoresistive transducers (Brown and Shaw, 1984), micro-indentation transducers (Ahmed and Burke, 1983), Fuji pressure-sensitive film (Liggins and Finlay, 1992; Zdero et al., 2001), ultrasound (Zdero et al., 2001), and X-rays (Kettlekamp and Jacobs, 1972). Due to the variety of loading conditions and measurement methods, there are slight variations in the measured tibiofemoral contact areas.

Three studies have provided details on the medial and lateral contact areas. Kettlekamp and Jacobs (1972) reported a mean contact area with the knee in full extension of 4.68 cm^2 on the medial plateau and 2.97 cm^2 on the lateral plateau. The contact areas reported by Walker and Hajek (1972) were noticeably lower. At full extension, the mean contact area was 2.22 cm^2 (medial) and 1.43 cm^2 (lateral). Fukubayashi and Kurosawa (1980) reported a medial contact area of 3.0 cm^2 and a lateral contact area of 2.3 cm^2 . The results of the reported contact areas are summarized in Table 6.1.

Table 6.1: The medial and lateral contact areas on the tibiofemoral joint (with menisci removed) reported in three studies. The amount of loading placed on the joint for the associated contact areas is also provided.

Study	Medial contact area (cm^2)	Lateral contact area (cm^2)	Magnitude of loading (N)
Kettlekamp and Jacobs (1972)	4.68 (2.50 – 6.73)	2.97 (1.74 – 5.11)	30 – 80
Walker and Hajek (1972)	2.22 ± 0.36	1.43 ± 0.29	1500
Fukubayashi and Kurosawa (1980)	3.0 ± 0.8	2.3 ± 0.8	1000

It would appear from Table 6.1 that as loading increases, contact areas decrease. However, the opposite trend has been found in two studies. Brown and Shaw (1984) and Fukubayashi and Kurosawa (1980) have determined that contact area increases as the applied load increases. Kettlekamp and Jacobs (1972) used an applied load of 30 to 80 N; Walker and Hajek (1972) used an applied load of 1500N; and Fukubayashi and Kurosawa applied loads between 200 and 1500 N (for the contact areas listed above, a load of 1000N was used). In light of this trend, the contact areas of Kettlekamp and Jacobs (1972) would be expected to be larger than those of Walker and Hajek (1972). Differences between the knee specimens themselves are not expected to affect the contact areas greatly, as no significant relationships were found between contact area and weight or sex and none between the ratio of contact areas and height, weight, or sex (Kettlekamp and Jacobs, 1972). The age and ethnic origins of the specimens have been offered as explanations for the differences, though, to the author's knowledge, their specific effects have not been reported. Therefore, based on the available information, the differences are may be partly due to the test methods and partly due to the specimens themselves. The loading contact areas of the FE models of this study were determined to be 343.5 mm² in the medial plateau and 214.5 mm² in the lateral plateau. These contact areas are within the range reported by Fukubayashi and Kurosawa (1980) and Kettlekamp and Jacobs (1972), and therefore are considered physiologically representative.

Despite the quantitative differences of the reported contact areas, it was consistently reported that the medial contact area was larger than the lateral contact area. The medial area of contact was reported to be approximately 1.6 times greater than the lateral through the first 35 degrees of flexion by Kettlekamp and Jacobs (1972). Walker and Hajek (1972) and Fukubayashi and Kurosawa (1980) found the medial area to be greater than the lateral by 30% and 25%, respectively. Harrington (1976) found a loading ratio of approximately 55:45 between the medial and lateral condyles. This occurrence was also found in the physiological gait cycle. During the stance phase of gait, a greater portion of the load was transmitted by the medial condyles (Morrison, 1970). Very few FE models have considered the physiological difference in loading of the medial and lateral condyles. One of the few FE studies that has considered separate medial and lateral loading is that of Little et al. (1986). The effects of separate loading (specifically,

the 55:45 ratio reported by Harrington (1976)) were found to be minor: a maximum stress difference of 0.2 MPa was reported. Therefore, loading was assumed evenly distributed over the medial and lateral condyles in the FE models of this study.

The forces transmitted in the knee joint during walking have been examined by Morrison (1970) and Harrington (1976). Morrison (1970) reported a maximum joint force in the range of 2 to 4 times body weight (with an average of 3.03 times body weight). Similarly, Harrington (1976) reported a maximum bearing force of 3.5 times body weight. The maximum joint forces were directed along the long axis of the tibia. Shear forces acting on the joint in the medio-lateral direction were found to be minimal (at most 0.26 times body weight) (Morrison, 1970). Other investigators (Askew and Lewis, 1981) have also assumed the effects of transverse shear forces to be minimal and have not considered it in their models. The stresses resulting from transverse shear forces are expected to be similar to those of moment loading (Askew and Lewis, 1981). The maximum force of 3 times body weight occurred during the stance phase of the gait cycle, approximately when the knee was in full extension in the single leg support phase. Harrington (1976) found the peak force occurred earlier in the stance phase compared to the data presented by Morrison (1970). A loading magnitude of 3 times body weight is generally accepted as being physiologically representative and has been used in numerous FE knee studies (e.g., Little et al., 1986; Hashemi and Shirazi-Adl, 2000; Nambu et al., 1991) as well as this study.

Forces transmitted by the knee in daily physiological activities have been studied by numerous authors. Morrison (1969) and Costigan et al. (2002) examined knee joint forces of stair climbing. Both studies reported an increase in joint force of 4 to 5 times body weight during stair ascension. Seedhom and Terayama (1976) reported that, when getting out of a chair without the aid of arms, the tibiofemoral forces were comparable with those occurring during walking. This information can be implemented for future applications of this FE tool; it has been presented here to provide a complete discussion of physiological loading.

6.2 Implementation of Near Physiologically Representative Load Conditions

For demonstration purposes, a loading case modeling the knee in full extension was used; any other loading can be easily incorporated following the methodology proposed below. A compression force of approximately three times body weight (Morrison, 1970; Harrington, 1976) of a 70 kg person, 2000 N, was non-uniformly distributed over the nodes at the surface of the subchondral bone. The load was distributed so as to approximate the contact area and corresponding pressure distribution patterns found by Fukubayashi and Kurosawa (1980). Such effort to represent the intrinsic loading conditions is rarely done in FE models.

The contact pattern was implemented into the FE models by digitally scanning the pattern from the published images (Fukubayashi and Kurosawa, 1980). The image was imported into Scion Image Analysis Program (Scion Corporation, Frederick, MD) and matched with the geometry of the tibia FE model (Fig 6.1). The borders of the 2, 4, 6 and 8 MPa pressure regions were determined for the FE model. The approximate contact areas in the FE model were determined to be 343.5 mm² for the medial region and 214.5 mm² for the lateral region. These areas fell within the ranges reported in Fukubayashi and Kurosawa (1980) and Kettlekamp and Jacobs (1972) (Table 6.1).

An in-house program was written using MATLAB[®] (Mathworks, Natick, MA) which determined the nodes situated within each of the borders (see Appendix A.3). A list of the nodes on the subchondral surface (and their corresponding coordinates) was first obtained. The list was formatted for input into MATLAB[®]. The in-house program sorted through the list to determine the region each node was situated in. Each node was then assigned a nodal force depending on the region it was situated in.

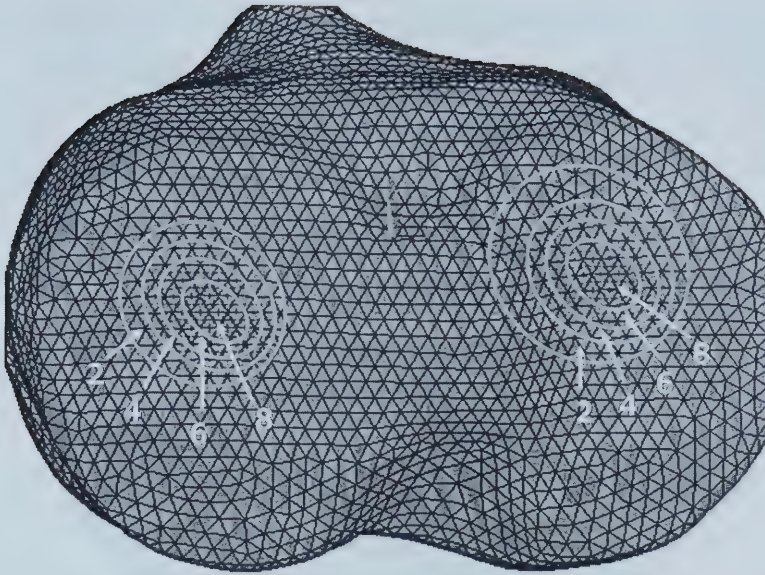


Figure 6.1: The tibia FE model showing the borders of the 2, 4, 6, and 8 MPa pressure regions, based on experimental findings. Nodes were assigned different loading depending on the region, allowing for a more representative distributed loading pattern.

Using the borders of the 2, 4, 6, and 8 MPa pressure regions, the areas occupied by each of these pressure regions were calculated. In the lateral region, the areas occupied by the 2, 4, 6 and 8 MPa pressure regions were 102.5, 62, 37.5, and 12.5 mm², respectively. A representative force was calculated for each region by multiplying the pressure by the contact area (e.g. 2 MPa multiplied by 102.5 mm² resulted in a representative force of 205 N). Therefore, 4 representative forces were determined for both the medial and lateral contact areas. A relationship between each set of resultant forces was found, resulting in a distribution of forces within each region. The sum of all 8 resultant forces from the medial and lateral regions corresponds to the total force on one leg (i.e. 2000 N). An example of the non-uniform load distribution in the tibia FE model is shown in Fig 6.2. The 2000 N resultant force was assumed to be parallel to the long axis of the tibia. Physiologically, the femoral long axis contacts to the tibia at 5 to 10 degrees valgus to the tibial long axis (Igbigbi and Msamati, 2002). The experimentally determined maximum joint force of 3 times body weight was directed along the tibial long axis; the shear force on the joint was found to be minimal (Morrison, 1970; Harrington, 1976). It should be noted that only loading from direct tibiofemoral contact was considered in the FE models of this study. Loading from soft tissues will be

considered in future FE models. Nevertheless, the currently implemented loading condition (i.e. 3 times body weight directed along the tibial long axis) is considered to be acceptable.

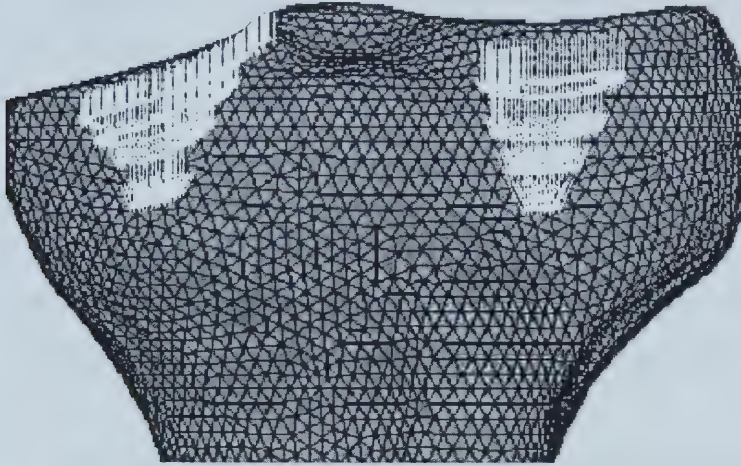


Figure 6.2: Non-uniform load distribution on the tibia FE model. The anterior surface of the tibia is shown.

The contact pattern assigned to the tibia FE model of this study was based on the pattern found using a 1500 N load (Fukubayashi and Kurosawa, 1980). It has been shown that increasing the loading also increases the contact area (Brown and Shaw, 1984; Fukubayashi and Kurosawa, 1980). However, the rate of change in contact area becomes smaller as the magnitude of the loads is increased; changing the load from 1000 N to 1500 N increased contact area by only 40 mm² (an 8% increase) (Fukubayashi and Kurosawa, 1980). Due to the decreasing rate of change, as the load is increased above 1500 N, the increase in contact area is expected to be even less. Therefore, it is reasonable to assume the contact area will remain constant for forces over 1500 N. However, as applied force increases over the same contact area, the pressure will increase. The change in pressure has been accounted for by the methodology described above. The 2000 N force and its associated contact pattern as applied to the FE models of this model can be considered physiologically representative.

The resultant contact force between the femur and the tibia are identical based on Newton's 3rd Law. In the FE models of this study, the long axes of the femur and tibia

are assumed to be parallel. As such, the forces transmitted between the femur and tibia are assumed to be identical in magnitude but opposite in direction. Strictly speaking, the femoral long axis is slightly valgus to the tibial long axis (Igbigbi and Msamati, 2002) and therefore shear forces are expected to be present. The effect of the shear forces will be investigated and implemented in future FE models. As stated earlier, soft tissues and menisci are not considered in these models. As menisci are not considered, the contact pattern between the distal femur and the proximal tibia must also be assumed identical. Therefore, the contact pattern implemented into the femur FE model was identical to that of the tibia FE model. Within these conditions, the loading used in the femur FE model would be more representative of the population with meniscectomized knees. The force distribution for the femur model was determined using a modification of the in-house MATLAB[®] program used for the tibia. An example of the non-uniform load distribution in the femur FE model is shown in Fig 6.3.

The distal end of the tibia and the proximal end of the femur was fixed in all directions, representing a rigid boundary condition. As can be seen in the sensitivity analysis in Section 3.2, this had a minor effect on the stress distributions in the region of interest. The diaphysis of the bones is fully constrained in all the FE models cited in this study. Nambu et al. (1991) performed a sensitivity analysis on the effect of constraints and found they had little effect on strains and displacements as long as the region of interest was far away from the base. As this study focuses on the proximal tibia and distal femur, fully constraining the mid-femur and mid-tibia is assumed acceptable.

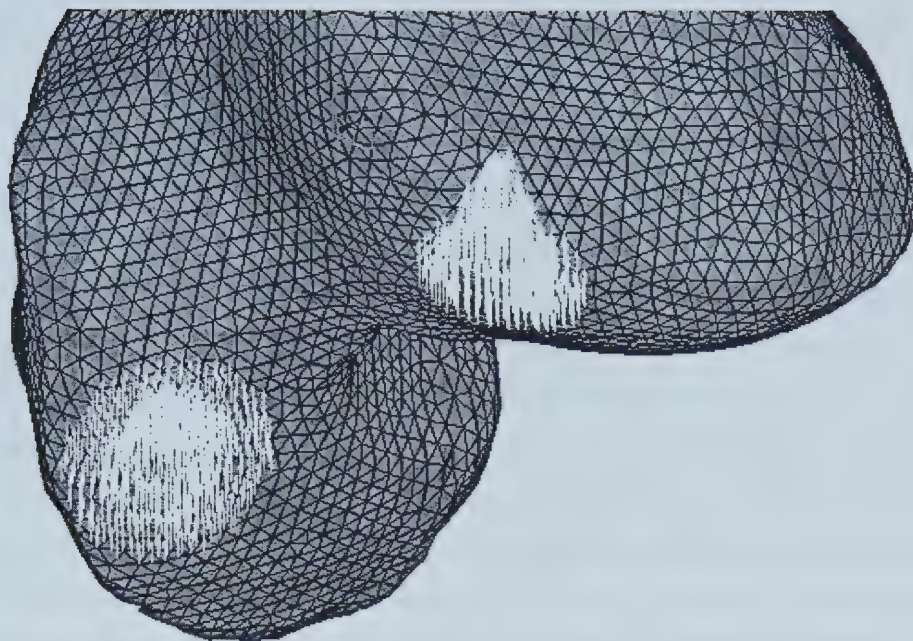


Figure 6.3: Non-uniform load distribution on the femur FE model. The condylar surface of the distal femur is shown.

Chapter 7: Examination of Finite Element Model Solutions

In the previous chapters, the geometry, material properties, and loading implemented in the finite element model were discussed. In this chapter, the solutions predicted by the models will be closely examined. The accuracy and validity of the results will be discussed. Accuracy refers to the ability of the mesh to approximate the exact solution; validity refers to the ability of the model to mimic the real structure. The accuracy of the model will be examined with several global and local accuracy checks. An interface analysis will be performed to ensure stresses are continuously transferred across the different bone volumes in the model, providing an accurate model. The validity of the model will also be evaluated by comparing the finite element results with those determined experimentally. As this tool can easily incorporate different sets of material properties, the effects of incorporating orthotropic and inhomogeneous bone properties are examined.

7.1 Interface Analysis

It is important to ensure there is no discontinuity of stress between adjacent bone volumes having different material properties (e.g. as ostensibly seen in the von Mises contour plot of Fig 7.1). Discontinuity will result in erroneous displacement (and therefore stress) predictions in the model. An investigation was done using a tibia FE model meshed with 2 mm elements. Distributed 2000N loading was used and a rigid boundary at the tibial base assumed. Von Mises stresses were recorded on paths plotted across the interfaces between volumes 5 and 8 (path I); volumes 16 and 17 (path II); volumes 20 and 21 (path III); volumes 4 and 5 (path IV); and volumes 11 and 19 (path V). The above volume numbers refer to the bone divisions in Fig 4.4. The five paths are shown in Fig 7.2, where von Mises stresses were recorded at the highlighted points.

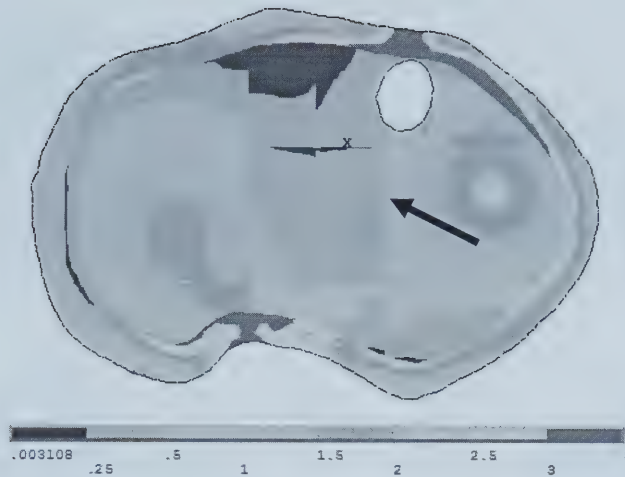


Figure 7.1: Von Mises contour plot at 20 mm below the joint line. Note the apparent stress discontinuity marked by the arrow.

The stresses were found to be continuous across adjacent regions of cancellous bone containing different material properties (Fig 7.3). The changes in stress were greater for the bone in the proximal layer as shown by paths I and IV. While these stress differences can be as large as 2.5 MPa as shown by path I, the transition of stresses across regions was found to be smooth. Although paths I and IV displayed greater changes in stress across interfaces compared to paths II, III, and V, the changes in stress remained smooth and continuous across all interfaces.

The interface between the cancellous and cortical bone sections was also examined. Two separate paths across the interface were examined; one of the paths is shown in Fig 7.4. As expected, there were much higher stresses in the very stiff cortical bone compared to the stresses in the cancellous bone (Fig 7.5). The data show, however, that the stresses from the two regions approached at the interface, as the cortical bone stresses decreased and the cancellous bone stresses increased about the interface plane. This confirms that, even with the large difference in stiffness, there is no discontinuity at the cancellous-cortical bone interface.

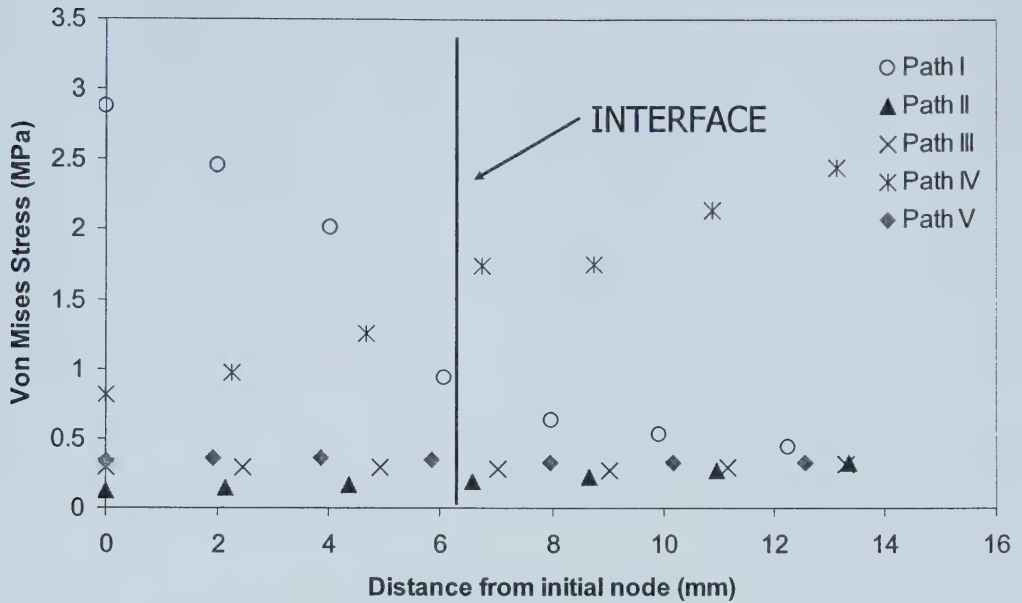


Figure 7.3: Von Mises stress curves of the five different paths used to examine the interface between adjacent regions of cancellous bone with different material properties in tibia FE model. The vertical bar denotes the approximate location of the interface between each pair of volumes. Von Mises stresses were recorded on paths plotted across the interfaces between volumes 5 and 8 (path I); volumes 16 and 17 (path II); volumes 20 and 21 (path III); volumes 4 and 5 (path IV); and volumes 11 and 19 (path V). See Figure 4.4 and Table 4.6 for volume and material property information, respectively.

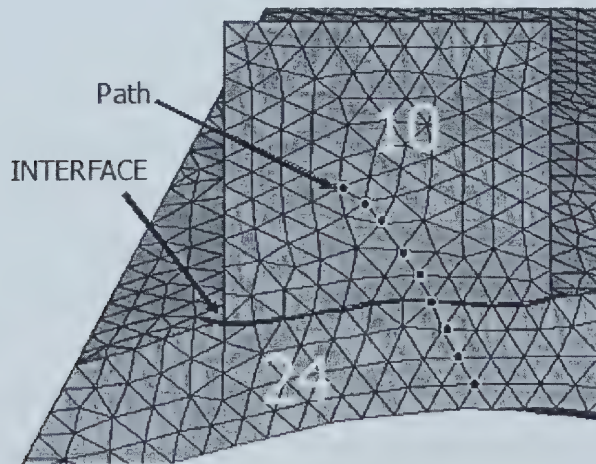


Figure 7.4: The nodal path plotted across adjacent cancellous and cortical bone regions in the tibia FE model. Von Mises stresses were observed at the highlighted points to examine the interface between the regions.

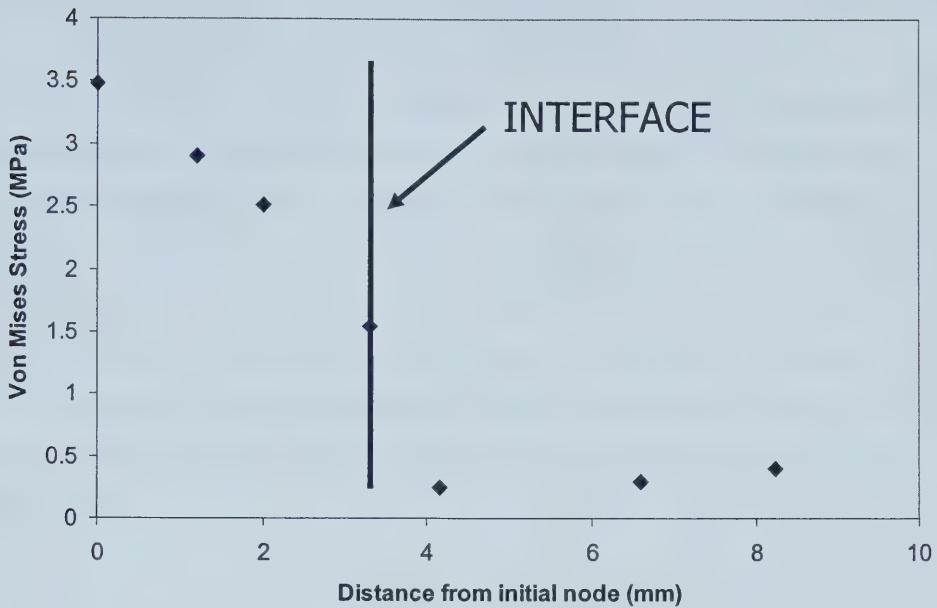


Figure 7.5: Von Mises stress curve of the path used to examine the interface between adjacent regions of cancellous and cortical bone in tibia FE model. The vertical bar denotes the approximate region for the interface between the volumes. Von Mises stresses were recorded on the path plotted across the interfaces between volumes 10 and 25. See Figure 4.4 and Table 4.6 for volume and material property information, respectively.

As all of the loading is placed on the subchondral bone, it is necessary to ensure that stresses are properly transferred from the subchondral bone to the cortical and cancellous bone. Interfaces between the subchondral and cortical bone, and subchondral and cancellous bone were both examined. The nodal path used to examine the stress continuity across the subchondral-cortical interface is shown in Fig 7.6. The subchondral bone has a relatively low stiffness and therefore its stresses are low. As the role of the subchondral bone is to transfer the stresses to the cortical bone (Hayes et al., 1976), it is expected that the cortex directly beneath the subchondral bone will have a high level of stress. This predicted trend is seen in the results (at the right side of the interface in Fig 7.7). Even with the large differences in the stiffness of the subchondral and cortical bones, stress continuity is still observed. The recorded stress at the interface node can be seen to connect the lower stresses of the subchondral region with the higher stresses of

the cortical region (Fig 7.7). This intermediate level of stress at the interface suggests there is a continuity of stresses across the subchondral-cortical bone interface.

The nodal path observed across the adjoining regions of subchondral and cancellous bone can be seen in Fig 7.8, and the corresponding stresses are shown in Fig 7.9. Because cancellous bone stiffness is not as high as that of cortical bone, the difference in stress levels between the subchondral and cancellous bone is not as great as the difference seen in subchondral and cortical bone (Fig 7.7). However, a similar pattern can be still seen at the interface between subchondral and cancellous bone. Lower stresses are transferred continuously from the subchondral bone to the higher stresses at the cancellous bone, as evidenced by the intermediate level stress at the interface node.

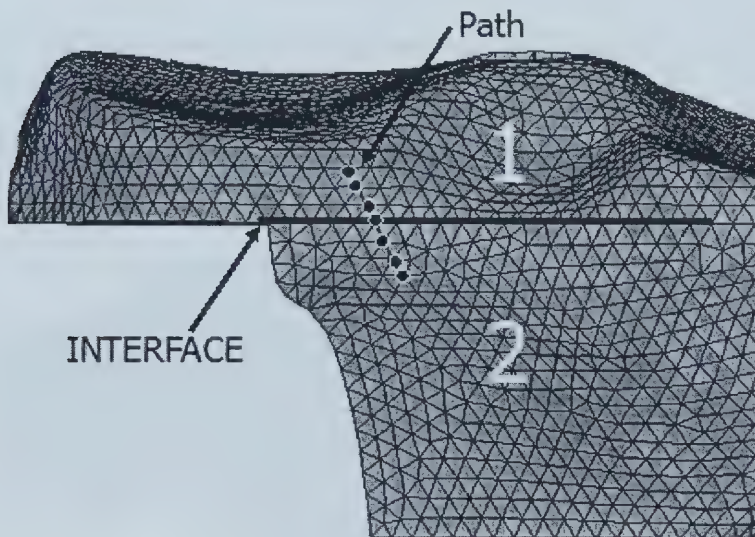


Figure 7.6: The nodal path plotted across adjacent subchondral and cortical bone regions in the tibia FE model. Von Mises stresses were observed at the highlighted points to examine the interface between the regions.

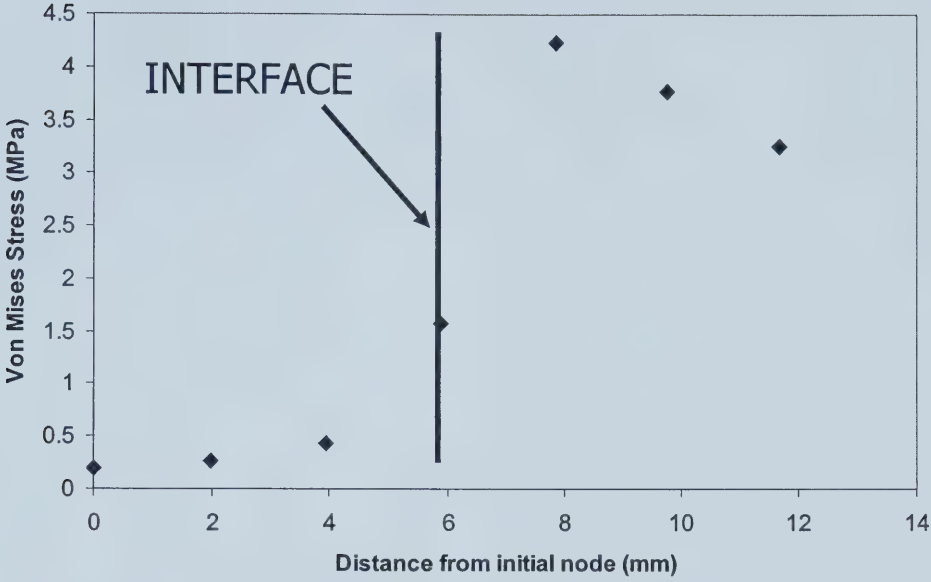


Figure 7.7: Von Mises stress curve of the path used to examine the interface between adjacent regions of subchondral and cortical bone in tibia FE model. The vertical bar denotes the approximate region for the interface between the volumes. Von Mises stresses were recorded on the path plotted across the interfaces between volumes 1 and 2. See Figure 4.4 and Table 4.6 for volume and material property information, respectively.

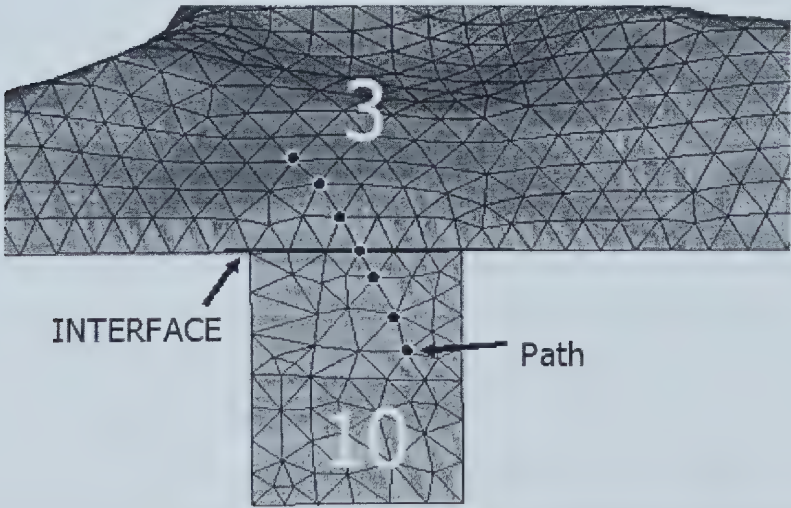


Figure 7.8: The nodal path plotted across adjacent subchondral and cancellous bone regions in the tibia FE model. Von Mises stresses were observed at the highlighted points to examine the interface between the regions.

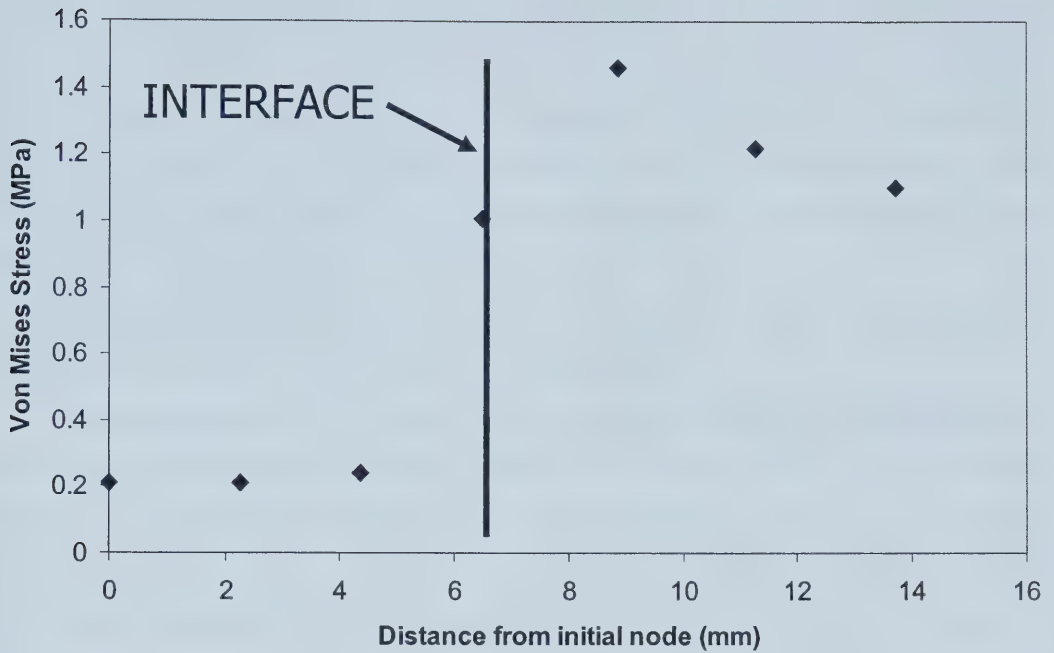


Figure 7.9: Von Mises stress curve of the path used to examine the interface between adjacent regions of subchondral and cancellous bone in tibia FE model. The vertical bar denotes the approximate region for the interface between the volumes. Von Mises stresses were recorded on the path plotted across the interfaces between volumes 3 and 10. See Figure 4.4 and Table 4.6 for volume and material property information, respectively.

A similar investigation of stress continuity for interfaces of the femur FE model was also performed. The model was loaded at two points in compression and a rigid boundary at the femoral base was assumed. The interfaces between cancellous bone volumes were examined using five nodal paths. Von Mises stresses were recorded on nodal paths crossing interfaces between volumes 7 and 11 (path i); volumes 16 and 17 (path ii); volumes 12 and 22 (path iii); volumes 9 and 11 (path iv); and volumes 7 and 16 (path v). The five paths are shown in Fig 7.10.

Continuous stresses were seen across all examined interfaces (Fig 7.11). Regions of higher stiffness can be seen to transfer stresses to regions of lower stresses. For example, in path ii of Fig 7.11, the stiffer bone (the left of the interface) experiences a rise in stress, while the less stiff bone (the right of the interface) experiences a similar trend. However, near the interface, stresses are continuous as can be seen with the

decline in stress. Similar stress continuity can be seen for the other four observed paths. A plot of nodal stresses across a cancellous-cortical bone interface (Fig 7.12) clearly showed a smooth increase of stresses from the cancellous bone (left side of interface, Fig 7.13) to the cortical bone (right side of interface, Fig 7.13). The distribution of the stresses from the subchondral bone to the cancellous and cortical bones was also examined. Stresses across the subchondral-cortical bone interface (Fig 7.14 and corresponding Fig 7.15) were clearly continuous as were those across the subchondral-cancellous bone interface (Fig 7.16 and corresponding Fig 7.17).

The interfaces examined in this analysis were only for one loading condition (i.e. a non-uniform 2000N distributed load). Under this load condition, the various interfaces examined are assumed to be representative of all the interfaces in the entire tibia and femur FE models. As the stresses were found to be continuous in all interfaces examined, it is expected that the stresses are continuous throughout the entire FE models. It is important to note that, while the stresses are continuous for this specific set of loading conditions, it does not preclude stress continuity across interfaces for other extreme loading conditions.

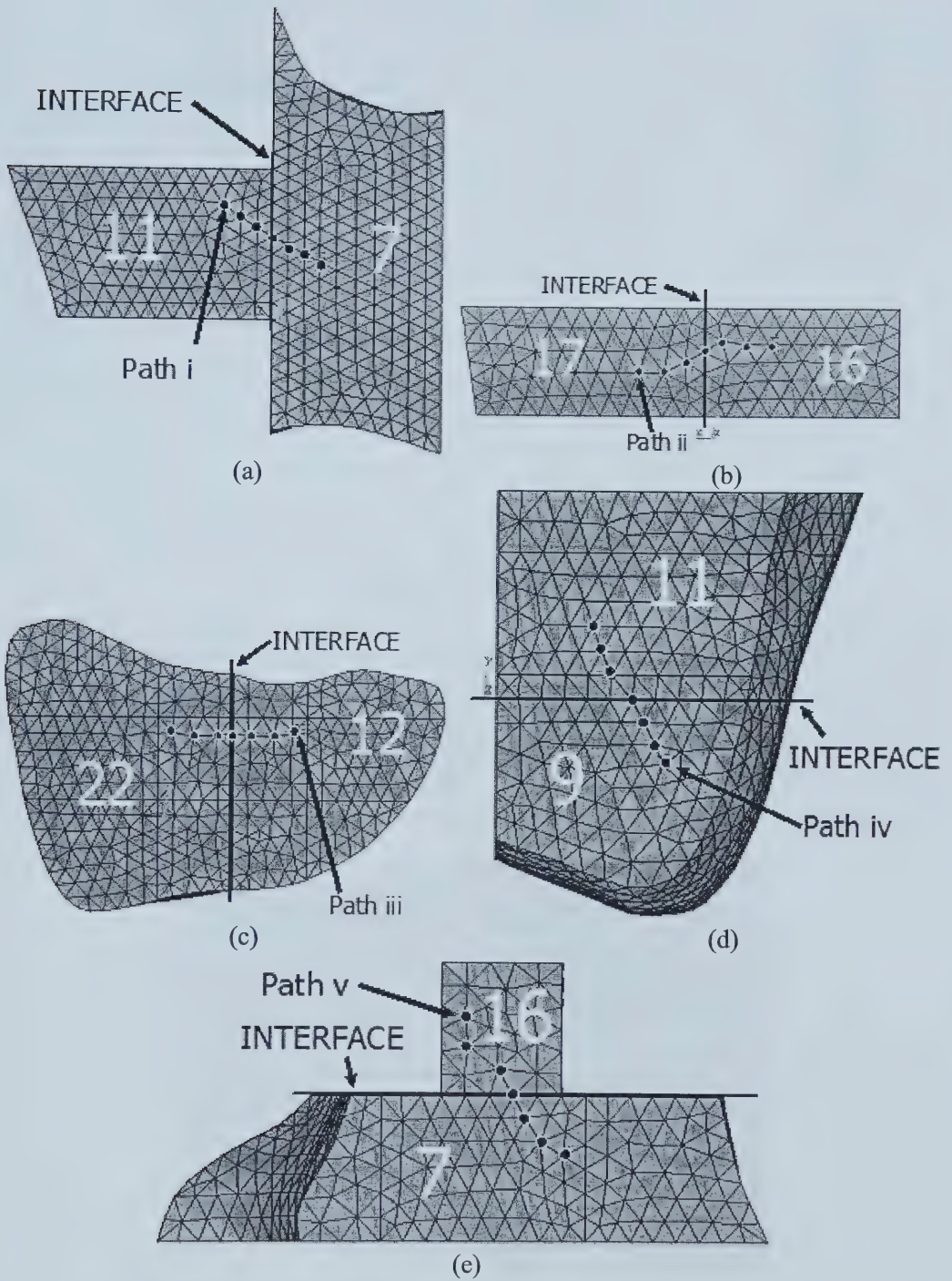


Figure 7.10: The five nodal paths plotted across adjacent cancellous bone regions with different material properties in the femur FE model. Von Mises stresses were observed at the highlighted points to examine the interface between the regions. Stresses from (a) path i, (b) path ii, (c) path iii, (d) path iv, and (e) path v are shown in Figure 7.11.

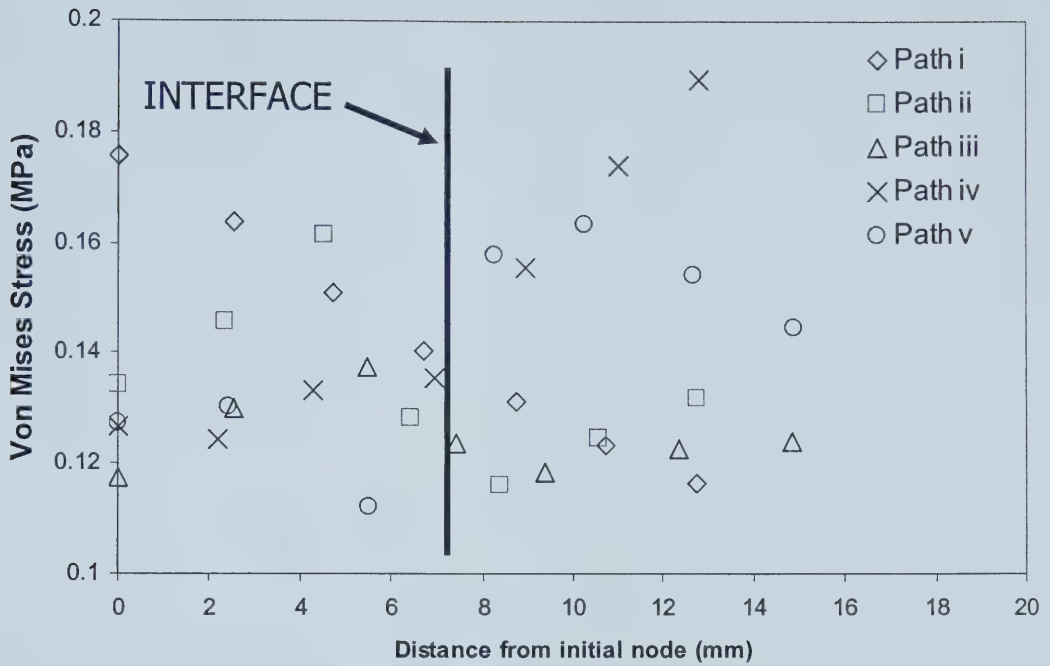


Figure 7.11: Von Mises stress curves of the five different paths used to examine the interface between adjacent regions of cancellous bone with different material properties in the femur FE model. The vertical bar denotes the approximate location for the interface between volume pairs. Von Mises stresses were recorded on paths plotted across the interfaces between volumes 7 and 11 (paths i); volumes 16 and 17 (path ii); volumes 12 and 22 (path iii); volumes 9 and 11 (path iv); and volumes 7 and 16 (path v). See Figure 4.5 and Table 4.7 for volume and material property information, respectively.

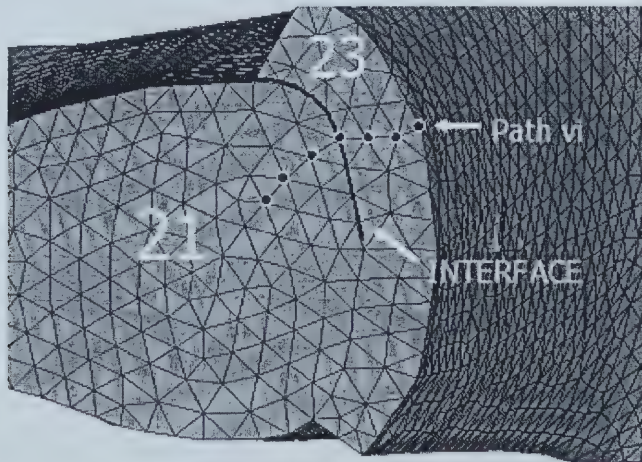


Figure 7.12: The nodal path plotted across adjacent cancellous and cortical bone regions in the femur FE model. Von Mises stresses were observed at the highlighted points to examine the interface between the regions.

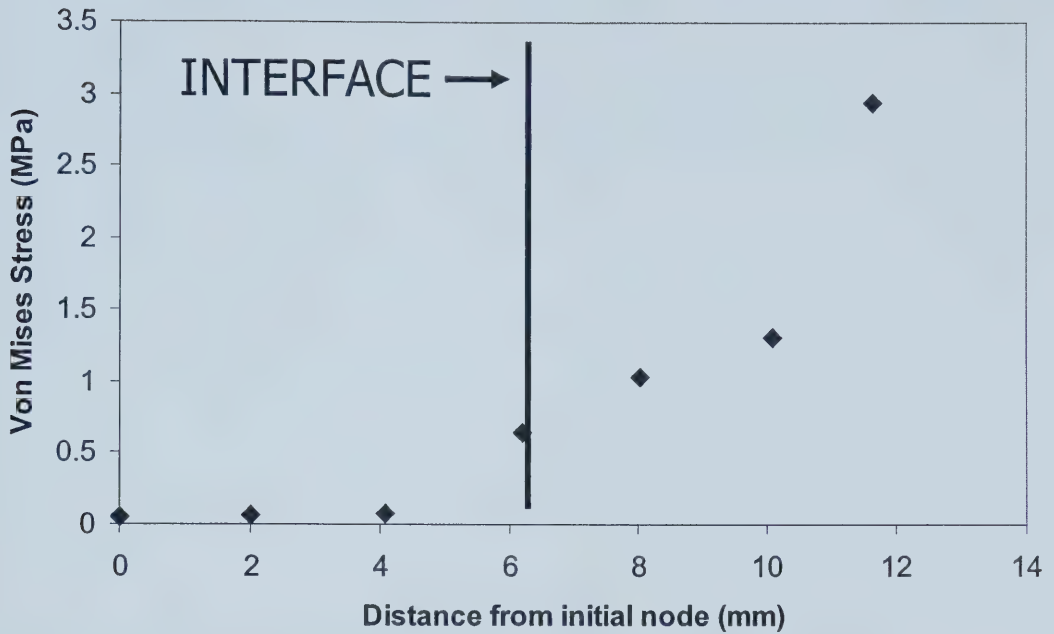


Figure 7.13: Von Mises stress curve of the path used to examine the interface between adjacent regions of cancellous and cortical bone in femur FE model. The vertical bar denotes the approximate location for the interface between the volumes. Von Mises stresses were recorded on the path plotted across the interfaces between volumes 11 and 23. See Figure 4.5 and Table 4.7 for volume and material property information, respectively.

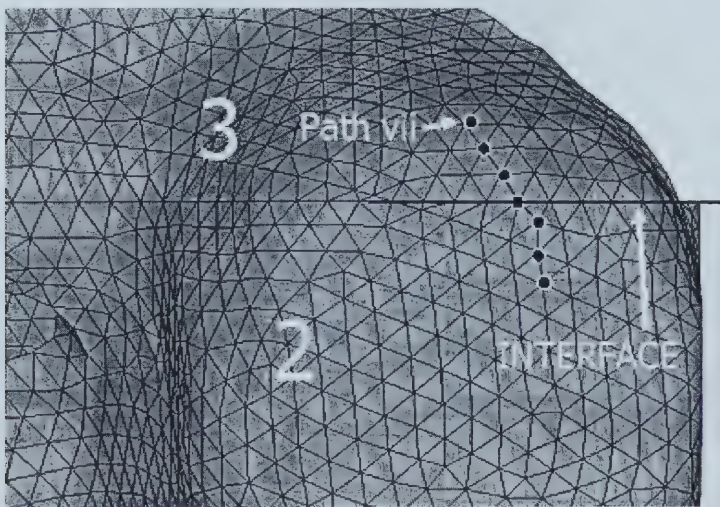


Figure 7.14: The nodal path plotted across adjacent subchondral and cortical bone regions in the femur FE model. Von Mises stresses were observed at the highlighted points to examine the interface between the regions.

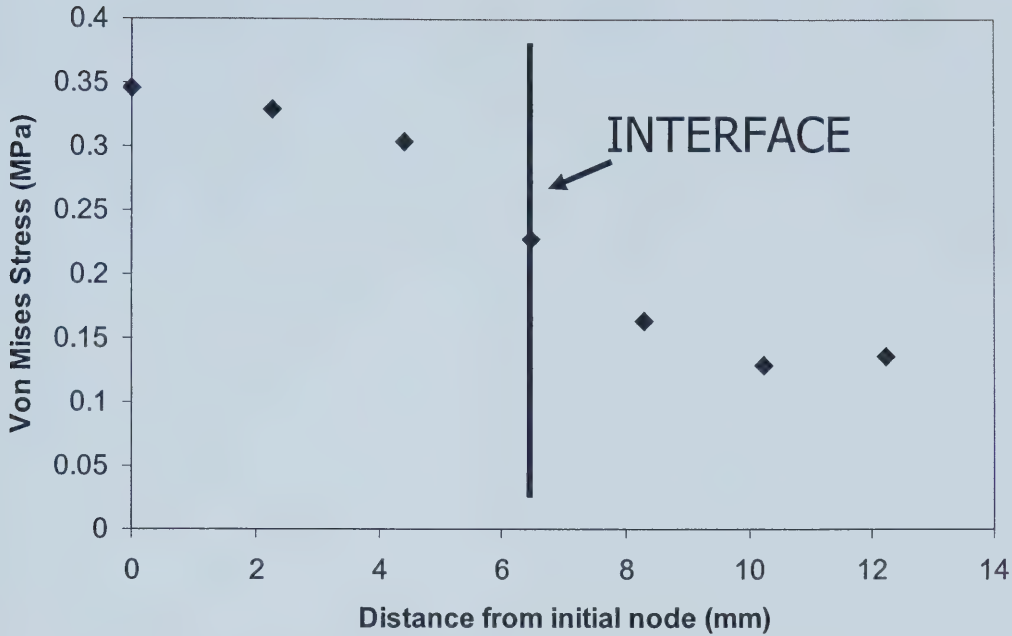


Figure 7.15: Von Mises stress curve of the path used to examine the interface between adjacent regions of subchondral and cortical bone in femur FE model. The vertical bar denotes the approximate location for the interface between the volumes. Von Mises stresses were recorded on the path plotted across the interfaces between volumes 2 and 3. See Figure 4.5 and Table 4.7 for volume and material property information, respectively.

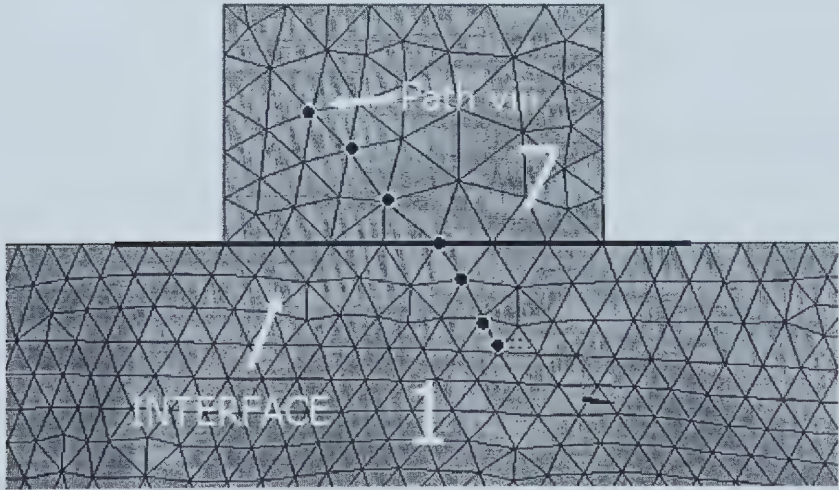


Figure 7.16: The nodal path plotted across adjacent subchondral and cancellous bone regions in the femur FE model. Von Mises stresses were observed at the highlighted points to examine the interface between the regions.

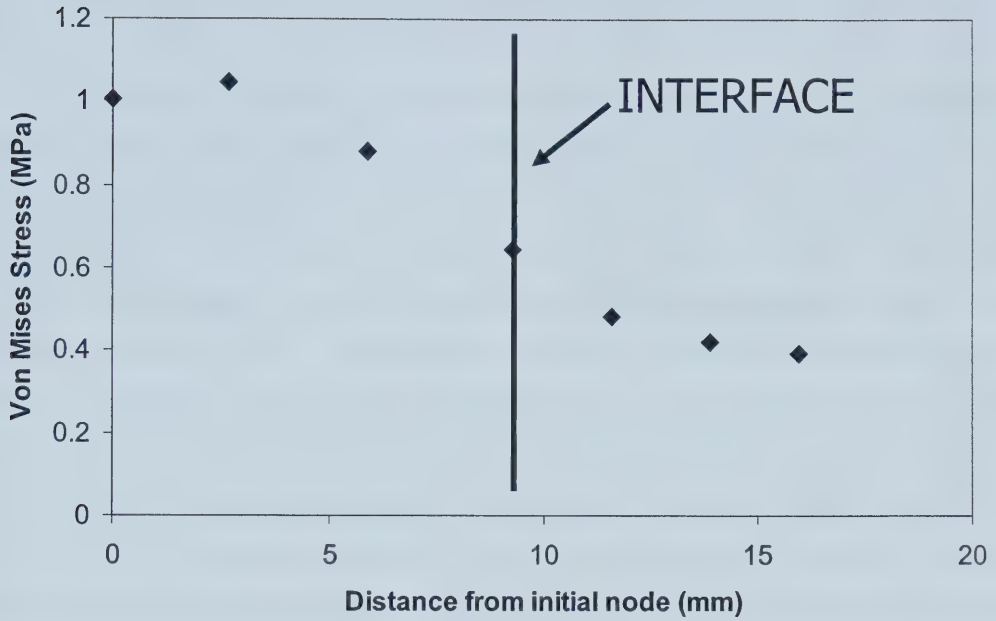


Figure 7.17: Von Mises stress curve of the path used to examine the interface between adjacent regions of subchondral and cancellous bone in femur FE model. The vertical bar denotes the approximate location for the interface between the volumes. Von Mises stresses were recorded on the path plotted across the interfaces between volumes 1 and 7. See Figure 4.5 and Table 4.7 for volume and material property information, respectively.

7.2 Effects of Incorporating Anisotropic and Inhomogeneous Bone

Cortical and cancellous bones of the femur and tibia are complex biological materials. Their anisotropy and heterogeneity must be considered to fully represent stress distributions. Cancellous bone within the tibia is highly inhomogeneous within individual tibiae and femora (Rho, 1992). Accurate characterization of the spatial variation in cancellous bone is necessary to accurately transfer stresses to the cortex (Askew and Lewis, 1981). The anisotropy has been underrepresented in FE models. While a combination of heterogeneity and anisotropy of the cancellous bone significantly affects resulting stresses in the bone (Askew and Lewis, 1981), it is not known what the precise effect of anisotropy in isolation is on the results (Lewis et al., 1982). This FE tool has the ability to easily incorporate different sets of material properties. As both anisotropic and isotropic bone properties are equally easy to incorporate, this tool could be used to investigate the effect of anisotropy. Furthermore, by applying detailed mapped bone data (Rho, 1992), the importance of characterizing the heterogeneity of bone upon stress distributions can be shown.

Two tibia FE models with bone tunnels were created. The bone tunnels represent a procedure performed during anterior cruciate ligament reconstruction surgery (see Chapter 8). Tunnels are created to allow placement and fixation of the graft replacing the ligament, and are a potential source of high stresses. The incorporation of anisotropic bone properties can allow for a more realistic portrayal of the forces surrounding the tunnel.

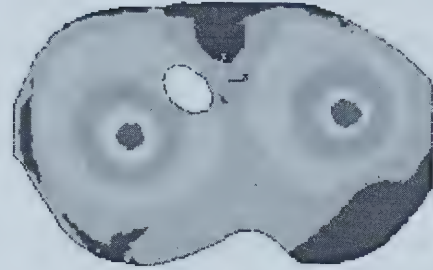
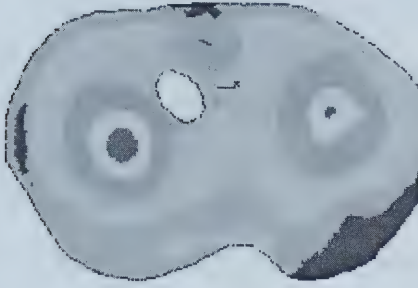
In one model, anisotropic and inhomogeneous bone properties (as described in Materials) were assigned; in a second model, isotropic and homogeneous bone properties were used. The Young's moduli of the isotropic bone were determined using a volume-weighted average of the axial moduli from Table 4.6. A Young's modulus of 19.9 GPa was used for the cortical bone, 837 MPa for the cancellous bone. A Poisson's ratio of 0.3 was used for both cortical and cancellous bone (Murase et al., 1983). Identical loading i.e. a 2000N distributed load was used for both models. A rigid boundary condition was used to constrain the base of the tibia.

Contour plots of von Mises stress at various transverse sections were compared. Major differences in stresses between the two models existed in the central cancellous bone and the cortex. At 15mm distal to the joint line (Fig 7.18a), higher stresses in the central cancellous bone as well as the cortical bone were predicted by the anisotropic-inhomogeneous model. The differences in stress also extended to the region around the bone tunnel. At 20mm below the joint line (Fig 7.18b), in the medial cancellous bone beneath the loading region, the isotropic-homogeneous model predicted a stress of 60% less than the corresponding region of the anisotropic-inhomogeneous model. In contrast, the stress levels were 100% greater in the cortex of the isotropic-homogeneous model than that of the anisotropic-inhomogeneous model. At sections more distant from the joint line (Fig 7.18c), the differences in stress level became larger. Because the cortical bone was stiffer than cancellous bone, the loads were transferred from the cancellous bone to the cortex thereby creating greater stress in the cortical bone. As the isotropic-homogeneous model had less stiff cancellous bone than the anisotropic-inhomogeneous model in the proximal tibia, higher stresses (up to 160%, Fig 7.18c) were predicted by the isotropic-homogeneous in the cortical bone and lower stresses (up to 50%, Fig 7.18c) were predicted in the cancellous bone.

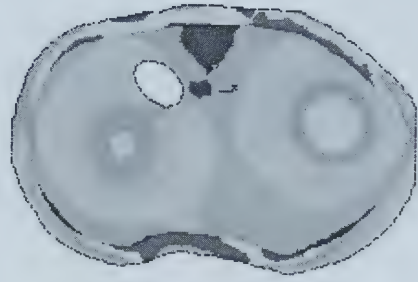
The spatial variation of the cancellous bone of the FE models allowed for a more realistic representation of the stresses in the bone. When the stiffer bone of the anterior and posterior regions was considered, higher stresses were detected in these regions as well as the surrounding cortical bone. When considering aspects such as prosthesis posts in the bone, stresses of cancellous bone surrounding the post are of particular importance. The use of isotropic and homogeneous material properties will not accurately depict the stress states at the bone-implant interface. The use of anisotropic and inhomogeneous bone allows this FE model to detect regions of low stress, which may be undetected otherwise.

Anisotropic & Inhomogeneous

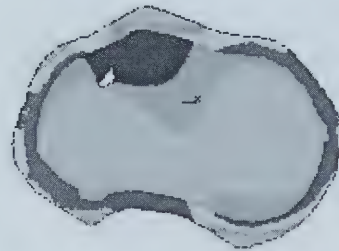
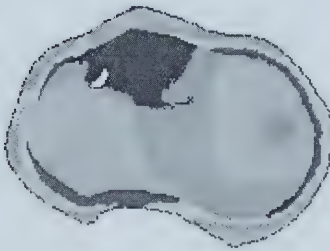
Isotropic & Homogeneous



(a)



(b)



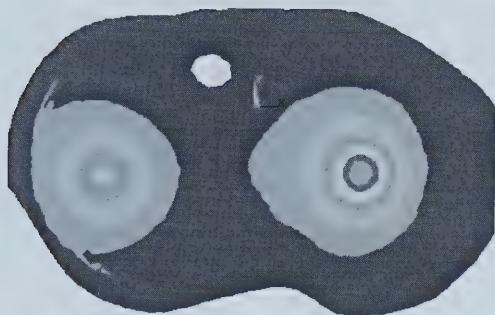
(c)



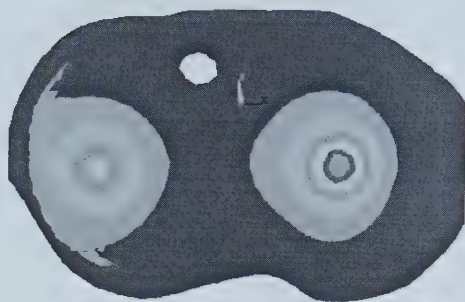
Figure 7.18: Von Mises stress contour plots of transverse sections of tibia FE model at (a) 15 mm, (b) 20 mm, (c) 30 mm below the joint line. The left-hand column shows plots of the model with anisotropic and inhomogeneous material properties; the right-hand column shows plots of the isotropic and homogeneous FE model. The bone tunnel (enclosed white region) has a circular cross-section but due to its orientation with respect to the transverse plane, the appearance of the tunnel cross-section evolves in each of the frames.

The individual effects of homogeneity and isotropy were each examined with the creation of two additional FE models. A FE model was created which incorporated isotropic and inhomogeneous bone properties, and a second model was created which incorporated anisotropic and homogeneous properties. The two models were loaded in identical manner to the above two cases. A comparison of the isotropic-inhomogeneous FE model to the anisotropic-inhomogeneous FE model showed that 100% lower stress levels were predicted in cancellous bone while 30% higher stress levels were predicted in cortical bone (Fig 7.19c). Stress levels were also found to be different between the anisotropic-homogeneous model and the anisotropic-inhomogeneous model (Fig 7.19). The cancellous bone stress levels were generally higher while the cortical bone stresses were lower. Incorporating isotropic bone may give incorrect predictions of highly stressed areas in the cortex. By incorporating homogenous bone, higher predictions of stress levels in the cancellous bone will result.

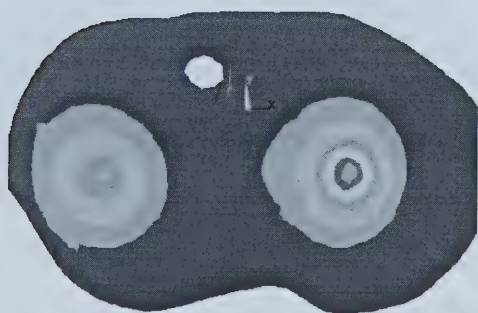
Anisotropic & Inhomogeneous



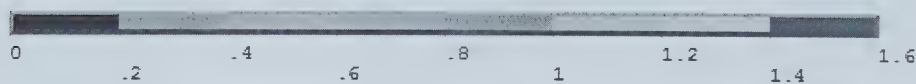
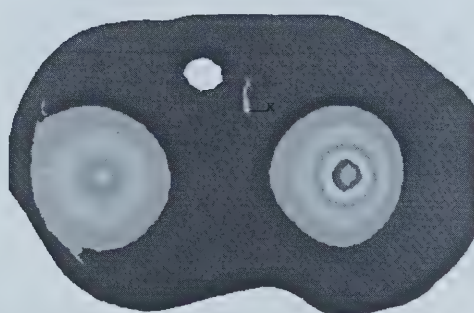
Anisotropic & Homogeneous



Isotropic & Inhomogeneous



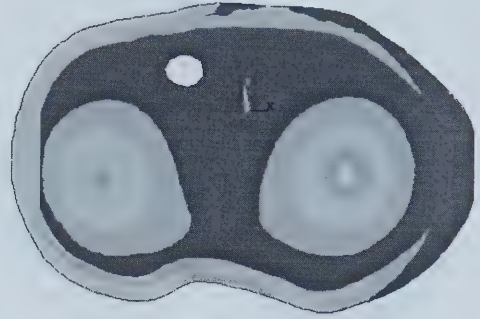
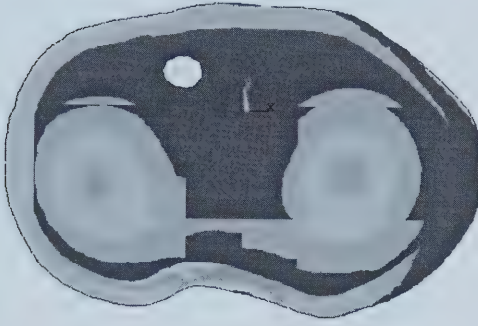
Isotropic & Homogeneous



(a)

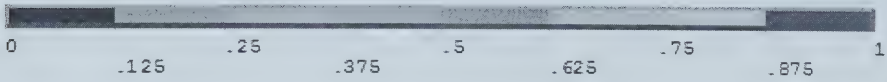
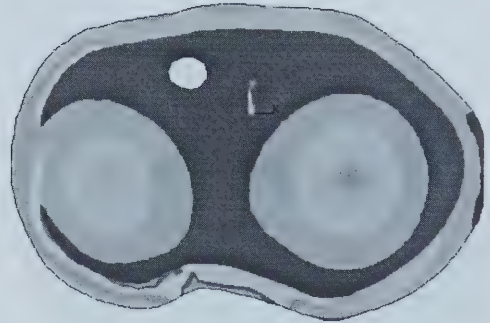
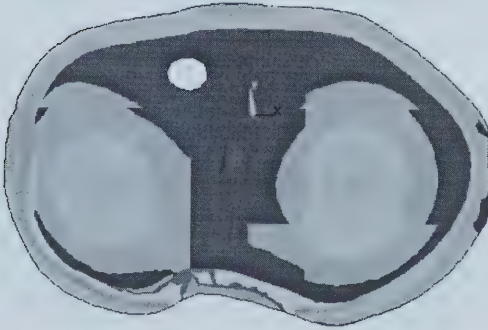
Anisotropic & Inhomogeneous

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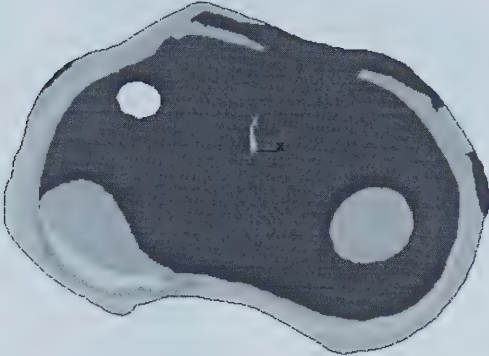
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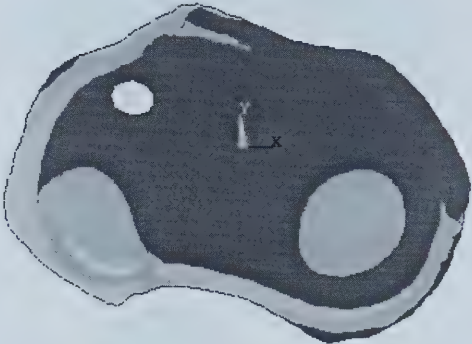


(b)

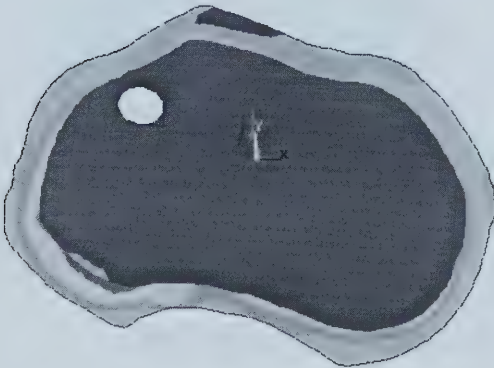
Anisotropic & Inhomogeneous



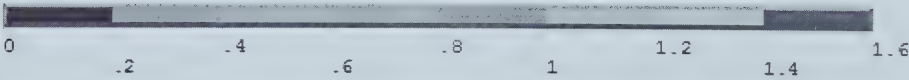
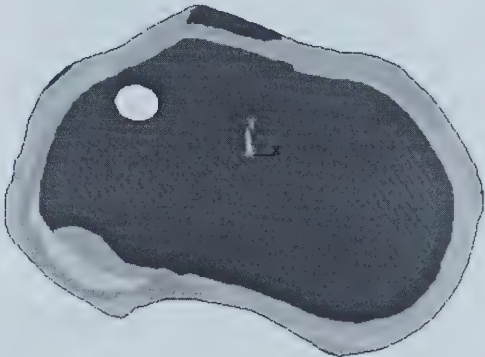
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Isotropic & Inhomogeneous



Isotropic & Homogeneous



(c)

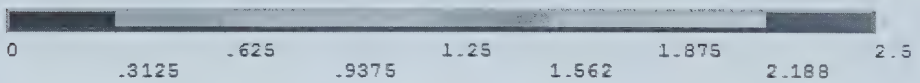
Anisotropic & Inhomogeneous

Anisotropic & Homogeneous



Isotropic & Inhomogeneous

Isotropic & Homogeneous



(d)

Figure 7.19: Von Mises stress contour plots of tibia FE models assigned anisotropic and inhomogeneous, anisotropic and homogeneous, isotropic and inhomogeneous, and isotropic and homogeneous material properties. Contour plots shows stresses at transverse sections (a) 15 mm, (b) 20 mm, (c) 30 mm, and (d) 40 mm below the joint line. The bone tunnel (enclosed white region) has a circular cross-section but due to its orientation with respect to the transverse plane, the appearance of the tunnel cross-section evolves in each of the frames.

7.3 Internal Validation

Although many investigations of stresses in bone have been performed using FE models, few investigators have addressed questions of modeling accuracy. FE models are often constructed without knowing the extent to which element size influences the stress results (Keyak et al., 1990). Thus, the reliability of those models is unknown. Analysis of the influence of element size is critical as numerical errors are inherent to FEM. The quality of the approximation can be determined using global and local accuracy checks. Local accuracy checks include internal force balances and stress convergence. Global accuracy checks include checking the deformation of the FE model and checking the smoothness of stress contours.

7.3.1 Internal Force Balance

Internal force balance checks on the tibia and femur FE models revealed there were negligible differences between the input force and the sum of the resultant nodal forces. A loading of 2000 N was evenly divided over two nodes (one on each condyle), and the base of each model was fully constrained. The resultant nodal forces of the models were summed and compared with the applied load. The difference in axial force was negligible for the tibia (1.25×10^{-10} N), and femur (6.36×10^{-11}) femur FE models. The sum of the resultant forces is negligible compared to the applied load. An internal force balance was performed only for the 2000 N non-uniform distributed loading because this load condition was used extensively for the FE models in this study. The results above do not preclude force balance failure at more extreme loading conditions.

7.3.2 Stress Convergence Analysis

A stress convergence analysis evaluating the von Mises stresses with increasing nodal degrees of freedom was performed for the femur and tibia FE models. The results can be seen in Chapter 5. Stresses converged at all six monitored locations on the tibia

model at an element size of 2 mm. Similarly, stresses converged at all eight monitored locations on the femur model at an element size of 2 mm.

7.3.3 Structural Deformation

Global error was checked qualitatively and quantitatively through the deformation of the overall femoral and tibial structures. A typical structural deformation of the femur FE model is shown in Fig 7.20. Note that the deformation is shown by the solid femur, with the mesh showing the non-deformed position. The compression loading placed on the condyles deforms the femur superiorly and posteriorly, which is the expected behaviour. These observations show qualitatively that the correct load direction was applied, and also that proper boundary conditions and constraints have been applied. The deformation of the tibia FE model was also verified in the same manner (Fig 7.21).

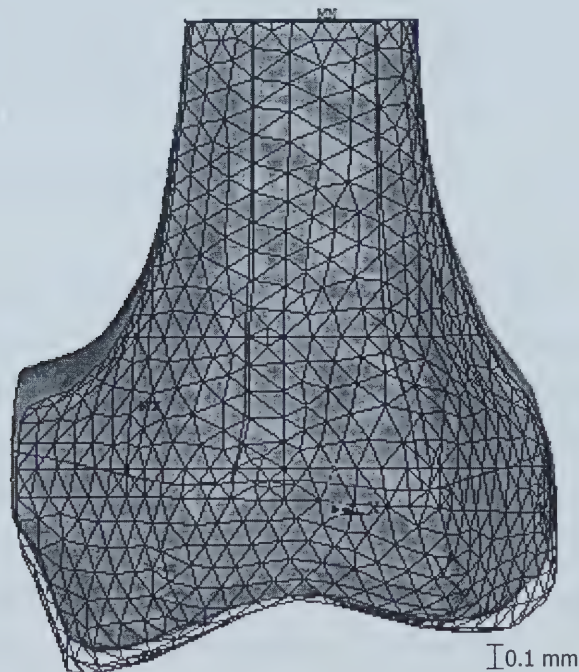


Figure 7.20: Structural deformation of the femur FE model upon compression loading of condylar surface. The mesh represents the original shape; the solid femur represents the deformed shape. The femur displaces superiorly, qualitatively verifying that correct loading and boundary conditions are being applied to the model.



Figure 7.21: Structural deformation of the tibia FE model upon compression loading of condylar surface. The mesh represents the original shape; the solid tibia represents the deformed shape. The tibia displaces posteriorly, qualitatively verifying that correct loading and boundary conditions are being applied to the model.

7.3.4 Displacement Convergence

A convergence analysis evaluating the displacements in three mutually perpendicular planes was performed for the femur and tibia FE models. Displacements were monitored with increasing nodal degrees of freedom. Convergence was found for all six monitored locations in the tibia model. The six observation points were identical to those used in the above stress convergence analysis. Convergence was also found at all eight monitored locations in the femur model. Again, these eight points were the same as those used in the femur stress convergence analysis above. The full details of the displacement convergence analysis were discussed in Chapter 5.

7.3.5 Strain Energy Error Convergence

For quasi-static linear elastic solids, strain energy is the energy stored in the system due to work done by external loads (Polgar et al., 2001). Global error and mesh adequacy was also verified through strain energy error. This method involves calculating the energy error within each finite element and expressing this error in terms of a global error energy norm (i.e. normalizing the global energy error in the model against the total energy). The result is a percent error in energy norm, which is a global estimate of mesh accuracy. The percentage error in energy norm is calculated by

$$J = 100 \left(\frac{e}{U + e} \right)^{1/2} \quad (7.1)$$

where J is the percentage error in energy norm, U is strain energy over the entire model (found by summing the strain energies of each element), and e is the energy error over the entire model, which is found by

$$e = \sum_{i=1}^{N_r} e_i \quad (7.2)$$

where N_r is the number of elements in the model.

Energy error convergence analyses were performed on both femur and tibia FE models. The energy error was observed separately for femur and tibia FE models meshed with 1.5, 2, 3, 4, and 5 mm tetrahedral elements. In addition to the overall error in the model, the error was determined separately for the cancellous bone, the cortical bone, and the subchondral bone. The change in the strain energy error for successive element sizes was calculated. In the femur FE model, the overall strain energy error was consistent throughout the 5 different meshes (Fig 7.22). The largest change in error was 3% between 3 and 4 mm. No change in strain energy error was found from 1.5 to 2 mm. The errors in all three bone types converged when the femur FE model was meshed with elements of 1.5 or 2 mm. Good convergence was obtained for the femur FE model with the errors all less than 5%. In the tibia FE model, the overall strain energy error was consistent throughout the 5 different meshes (Fig 7.23). The change was only 1% for mesh densities larger than 3 mm, with no change occurring at mesh densities of 1.5 mm and 2 mm. The greatest amount of change was found in the cancellous bone with a 4%

change between the 3 mm and 4 mm mesh. Therefore, good convergence of error in strain energy was found for the tibia FE model for all five mesh sizes.

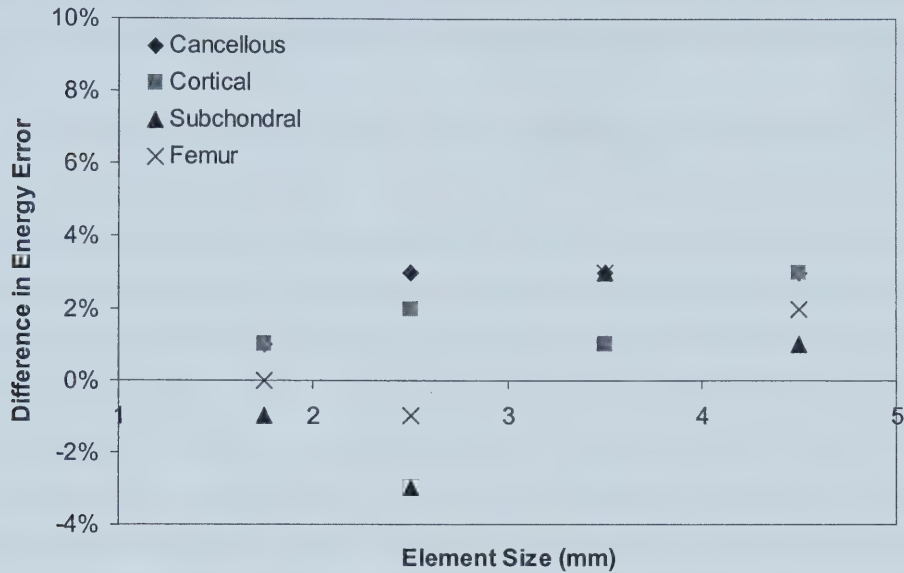


Figure 7.22: The change in strain energy error measured over successive element sizes for the femur FE model. Points located between 4 and 5 mm denote the change in energy error between 4 and 5 mm elements (and similarly for the other element sizes).

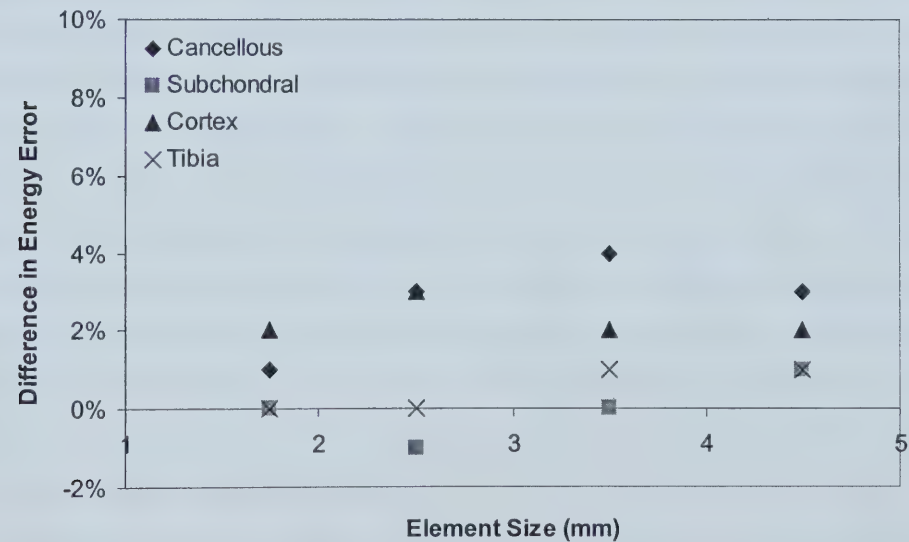


Figure 7.23: The change in strain energy error measured over successive element sizes for the tibia FE model. Points located between 4 and 5 mm denote the change in energy error between 4 and 5 mm elements (and similarly for the other element sizes).

Considering the variations in the input parameters from experimental sources (in particular, the large variations in the experimental data of material properties), the small changes in percentage errors of the strain energy of the FE model are comparatively small. The convergence of these small errors shows that this is an accurate FE model.

7.4 External Validation: Verification with Experimental Data

Very few published experimental stress data of intact human proximal tibiae and distal femora are available. Experimental *in vitro* analyses of intact human tibiae have been performed by Finlay et al. (1982), Bourne et al. (1984a, 1984b), Reilly et al. (1982), and Krause et al. (1981). These studies used strain gauges to provide information on the surface patterns of strain in normal physiological and varus/valgus deformed loading. Due to the lack of necessary details in other studies, the experimental results of Bourne et al. (1984a) and Reilly et al. (1982) were used for comparison with our model's results, but it is important to note that there are several differences between our model and the *in vitro* strain gauge experiments of Bourne et al. (1984a) and Reilly et al. (1982). The load distribution information used for the models was gathered from a different set of loading conditions (i.e. from Fukubayashi and Kurosawa, 1980), as experimental information was not provided (however, the total load was kept constant for comparison). Due to the inherent anatomical differences in geometry between tibiae used in experiments, the locations on the model used to determine strains were only approximately equivalent locations to the experimental strain gauge locations. Local discrepancies may also arise in our model since ligament forces were not considered. In addition, the bone properties assigned in the finite element models are approximations based upon experimental data obtained in a different study. Considering these differences, the comparison between numerical and experimental values should be seen as an order of magnitude analysis rather than a strict numerical comparison.

In the experiment by Bourne et al. (1984a), the strains were determined on six tibiae from 17-, 20-, and 33-year-old specimens with an applied load of 2453 N. Reilly et al. (1982) determined the strains on four specimens (the age of the specimens were not reported). The tibiae were loaded in compression with 2250 N.

As stiffness of bone is known to vary with age (Ding et al., 1997; McCalden et al., 1997; 1993, Burstein et al., 1976), the difference in bone age between the experiment (23 years, Bourne et al. (1984a)) and the model (60 years) was expected to be the main factor influencing the comparison. To account for this using the experimental findings of Burstein et al. (1976) (i.e. cortical bone stiffness decreases 4.7% per decade; shear modulus decreases 4.0% per decade), the 60-year-old bone cortical bone properties (used in our model) were extrapolated to those at age 20. Cancellous bone properties were extrapolated using apparent density relationships. Apparent densities of each cancellous bone volume were first determined using stiffness-apparent density relationships (Ashman et al., 1989) and extrapolated to 20 years of age using an apparent density-age trend (McCalden et al., 1997). The new apparent density was then used to determine the stiffnesses and shear moduli using the empirical relationships of Ashman et al. (1989).

A loading of 2453 N was non-uniformly distributed on the subchondral bone to reflect the load used in the experiment by Bourne et al. (1984a). The strain gauges used in the experiment measured approximately 6.2 x 7.6 mm. On the finite element model, nodal strains within a 6 x 7 mm region at each equivalent strain gauge location were averaged to obtain approximate strains. The values at 25 mm and 75 mm distally from the joint line were compared. The experimental principal strain values of 7 gauges are shown in Table 7.1 together with the principal strains as predicted by the model. The principal strain directions ranged from -10.5° to +19.6° as measured clockwise from the long axis of the tibia in the strain gauges. For the equivalent model locations, a range of -22.4° to 23.1° was found. The orientation of the principal strain angles was less agreeable than the principal strain magnitudes with the experimental data, and should be investigated further in the future. Except for locations 1 and 2 (at 25 mm below the joint line), the principal strains are in the same order of magnitude. At all 7 locations, the signs of the model and experimental principal strains match. Similar strain patterns at the distal region of the tibia were found between the *in vitro* and finite element models. High strain levels in the medial and posterior regions found by Bourne et al. (1984a) were also predicted by our models.

Our finite element model results were also compared with experimental results from Reilly et al. (1982). Six locations at 60 mm below the joint line of the tibia were

compared (Table 7.2). The strains between the finite element model and the experiments were of the same order of magnitude at four of the six locations. At these four locations, the pattern of higher medial and lateral strains seen in the *in vitro* test was similarly detected by our model.

Table 7.1: Comparison of nodal strains from finite element model with experimental strains from Bourne et al. (1984a). Locations of strain gauges are shown in parentheses: A = anterior, P = posterior, L = lateral, M = medial. The strain gauge numbers correspond to those appearing in Bourne et al. (1984a). ϵ_1 and ϵ_2 are principal strains and θ is the principal strain angle.

Strain Gauge Number	Distance from joint line (mm)	Finite Element Model Results			Experimental Results		
		ϵ_1	ϵ_2	θ	ϵ_1	ϵ_2	θ
1 (L)	25	-39	3	23.1	-110	120	3.7
2 (M)		-32	20	-22.4	-190	100	2.3
8 (L)	75	-105	44	-2.8	-100	40	19.6
9 (A)		-24	11	12.7	0	25	-6.9
10 (A-M)		-122	35	2.6	-60	20	-10.5
11 (M)		-222	84	-8.1	-180	50	-5.6
12 (P)		-365	205	9.3	-270	80	2.9

Table 7.2: Comparison of nodal strains from finite element model with experimental strains from Reilly et al. (1982) at 60 mm below the joint line of the tibia. Locations of strain gauges are shown in parentheses: A = anterior, P = posterior, L = lateral, M = medial.

Strain Gauge Location	Finite Element Model Results	Experimental Results
P-L	-349	-11
P-M	-362	-168
L	-227	-253
A-L	-6	-49
A-M	-177	-161
M	-280	-333

Chapter 8: Clinical Applications of Finite Element Tool

In this chapter, a demonstration of potential clinical applications of this tool is given. The flexibility of this tool in regards to modeling geometry, material properties, and loading conditions allows for creation of finite element models, which allow for a better representation of stress states over previous FE models, for numerous clinical applications. By modeling surgical alterations of the femur performed during anterior cruciate ligament reconstruction, an investigation of the stresses in the bone can be examined. In particular, the cortical stresses surrounding the femoral condylar aperture resulting from button-like fixation are investigated. In modeling surgically altered tibiae, cancellous bone stresses surrounding the bone tunnel are examined to determine an optimal tibial tunnel placement. A second application of the developed tool is for improvement of the design of total knee arthroplasty implants. Four different tibia prosthesis designs were compared using bone-implant interface forces and internal stress distributions.

8.1 Application of Finite Element Tool to ACL Reconstruction

A short explanation of the operative procedure to repair a damaged anterior cruciate ligament is given below. Typically, an ACL reconstruction is done arthroscopically and takes approximately 2 to 3 hours. First, a graft of either bone-patellar tendon-bone type or hamstring (semitendinosus/gracilis) type is obtained through an incision in the knee. Before insertion of the graft, the remaining portion of the torn ACL is shaved off (or debrided). Holes are drilled in the femur and tibia in order to place the graft. A hole is first drilled completely through the tibia from the outside, and then a blind-ended hole is drilled into the femur (a smaller guide hole is drilled beforehand to ensure proper tunnel placement). The graft is then pulled into the tunnels via sutures. It is then secured by placing interference screws at the bone blocks (if a bone-patellar tendon-bone graft is used) or anchored using a button (on the femoral end) if a hamstring graft is used. An illustration of a reconstructed ACL is shown in Fig 8.1.

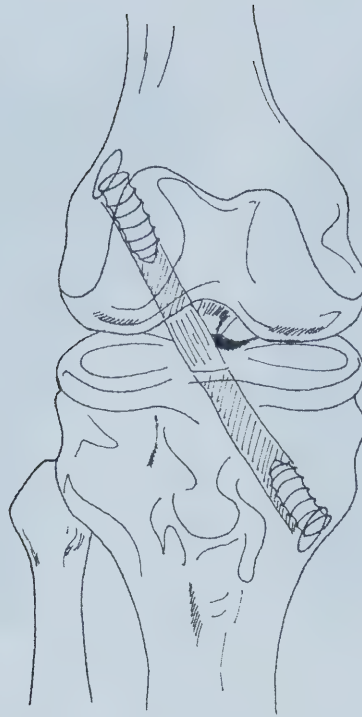


Figure 8.1: Schematic diagram of an anterior cruciate ligament reconstruction. Two interference screws are used to secure the graft in the bone tunnels.

8.1.1 Tunnel Placement

To examine the change in stresses created by the introduction of a bone tunnel, tibia FE models were created with and without representative bone tunnels. Each model was loaded with a distributed 2000 N load and rigid boundary conditions were assumed. The model did not implement the loading from the graft and fixation devices. The tibial tunnel was initiated at the midpoint between the anterior tibial tubercle and postero-medial border of the tibia, as in clinical practice (Fu et al., 2000). Bone tunnel positions were varied at 10, 15, 20, and 27 degrees relative the midsagittal plane (approximately 2 cm medial from the tibial tubercle) (Fig 8.2a), and at 25, 30, 35, and 40 degrees relative to the midcoronal plane (Fig 8.2b). The effect of increasing the tunnel diameter was also investigated using models with bone tunnel diameters of 6, 7, 8, and 9 mm. The angles and diameters are within the limits of those that may occur in clinical practice. The bone tunnels penetrate completely through the tibia and are approximately 5 cm in length.

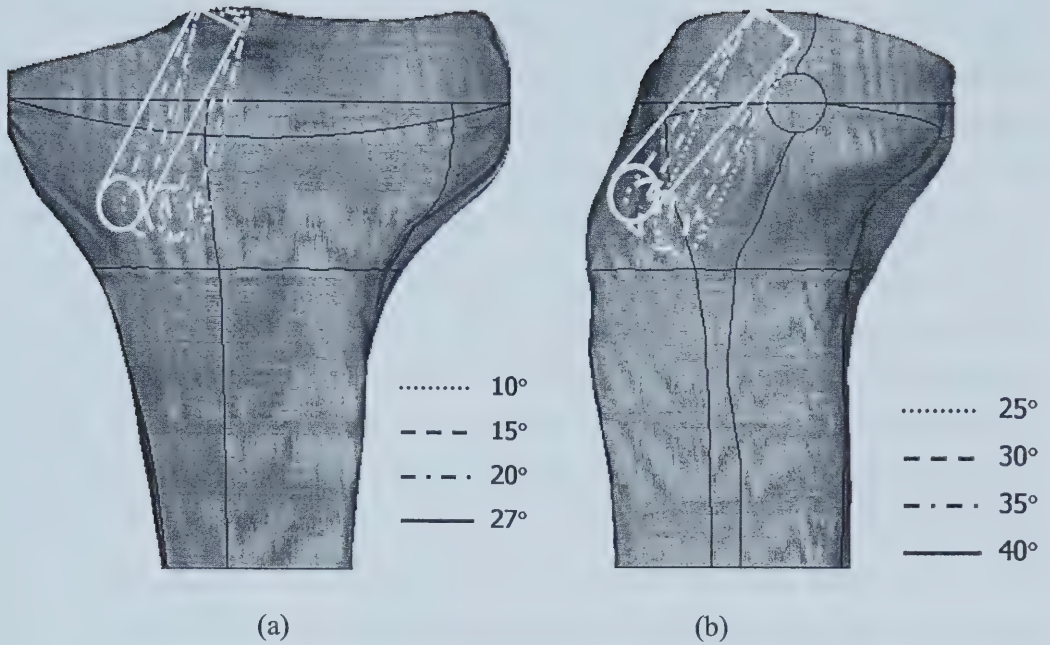


Figure 8.2: Schematic of bone tunnel placements in the tibia FE model. Tunnels were rotated (a) 10, 15, 20, and 27 degrees relative to the midsagittal plane and (b) 25, 30, 35, and 40 degrees relative to the midcoronal plane.

The introduction of a bone tunnel affects the internal stresses of surrounding bone. In the model, a bone tunnel was placed at 25 degrees anterior to the midcoronal plane and 10 degrees medial to the midsagittal plane. The introduction of the tunnel reduced stresses in the region of cortical bone at the distal tunnel aperture by approximately 70% (Figs 8.3d and j). At the proximal aperture region, stresses increased by as much as 80% with the addition of the tunnel. Regions of cortical bone slightly more proximal (20 – 25 mm from the joint line) to the aperture of the bone tunnel were less affected by the introduction of the tunnel. Stresses were observed to increase only 40% in these regions (Figs 8.3b and c). Cancellous bone in the regions at the posterolateral tunnel wall experienced a 50% reduction in stress along the complete length of the tunnel (Figs 8.3a to d). The region of cancellous bone near the tunnel aperture (at the cortical-cancellous bone interface) also decreased by as much as 50% (Fig 8.3j). Subchondral bone in the regions of the tunnel increased in stress by as much as 40% with the addition of the tunnel. While the tunnel was shown to create a region of altered stress

in the immediate surrounding bone, the influence of the tunnel does not affect most regions of the tibia. In regions where the bone tunnel does not penetrate, the stresses are relatively unchanged (Figs 8.3e to i). While the creation of bone tunnels in anterior cruciate ligament reconstruction is absolutely necessary, the transformation of internal stresses in regions of bone near the tunnels should not be ignored. Therefore, examination of the stress trends resulting from tunnel placement positions is necessary.

In comparing the stress contour plots with different tunnel positions, generally the lowest stresses occurred in the postero-lateral region surrounding the bone tunnel and the highest stresses occurred in the postero-medial regions. Tunnels which were placed less medially (and therefore closer to the bone at the midsagittal plane) showed lower tunnel wall stresses (e.g. a tunnel placed at 10 degrees relative to the sagittal plane (Fig 8.4a) compared to one placed at 27 degrees (Fig 8.4d)). In the case where the tunnel was placed at 10 degrees from the sagittal midline, the postero-lateral tunnel wall showed almost no stress. The incidence of lower stress on the lateral tunnel wall is expected as less stiff bone exists in the central region of the tibia. Similarly to the above, the posterior tunnel wall was exposed to lower stress as the tunnel was rotated less anteriorly (e.g. comparing a tunnel placed at 25 degrees relative to the coronal plane (Fig 8.5a) compared to placement at 40 degrees (Fig 8.5d)). Regardless of bone tunnel position, stresses were consistently lower in the region of bone at postero-lateral tunnel wall compared with the addition of a tunnel (c.f. Figs 8.4 and 8.5 to Fig 8.3). A reduced stress level in the bone should be avoided, as it will eventually lead to bone resorption. While a tunnel rotated closest to the midcoronal and midsagittal planes would reduce surrounding bone stresses, this position is not optimal. However, at the other extreme, tunnel placement with a large medial and anterior rotation will cause the bone to experience large stress increases. Cortical stresses at the antero-medial region were predicted to increase 600% with the addition of a tunnel rotated 27 degrees from the midsagittal plane (c.f. Fig 8.4d to 8.3b). Clinically, a tunnel is inserted at 1 to 2 cm medially from the tibial tubercle (Morgan et al., 1995). From the FE models, it is predicted a placement closer to 2 cm will produce greater internal stresses than at 1 cm; introducing the tunnel at 1.5 cm may reduce internal bone stresses. A typical tunnel is placed with a 35 to 45 degree rotation relative to the midcoronal plane (Ayerza et al., 2003). Within this range and this

set of loading conditions, the addition of a tunnel may cause cancellous bone at the postero-medial tunnel region to double in stress. However, cortical stress is only expected to increase by a maximum of 40% (c.f. Fig 8.5d to 8.3b). As the relative increase in cancellous bone stresses is not overly large, and to avoid possible bone resorption, placement of the tunnel at 45 degrees is recommended. Internal bone stresses resulting from tunnel placement for anterior cruciate ligament reconstruction has not been previously studied. Therefore, it is difficult to compare the results predicted here with other FE studies.

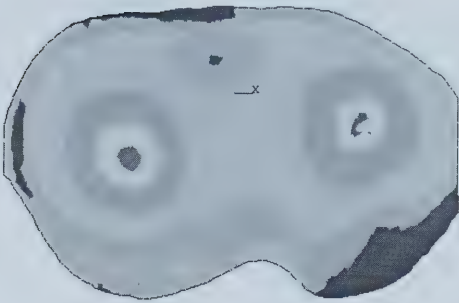
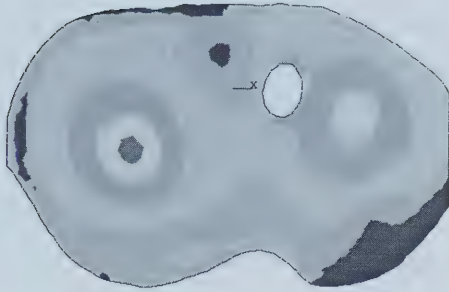
The internal stresses predicted in the tunnel enlargement simulations can represent the effects of intentional (i.e. from surgical procedure) or unintentional enlargement (e.g. reported post-operatively). Enlarging the tunnel diameter resulted in a slight increase in the postero-lateral tunnel wall stress. More evidently, the walls of the larger tunnels were closer to the cortical-cancellous bone interface resulting in an increase of anterior cortex stress. This trend was observed at all lengths of the tunnel, i.e. at 15 mm (Fig 8.6), 20 mm (Fig 8.7), and 25 mm (Fig 8.8) below the joint line. At the cortical aperture of the tunnel (at 30 mm below the joint line), the stresses in the cortical bone immediately surrounding the tunnel increased with tunnel enlargement (Fig 8.9). The cortical stress levels are considerably higher than those in the cancellous bone. While the levels of stress predicted from these simulations are not expected to cause bone fracture, a greater danger may exist in the reduction of stress levels caused by the introduction of a bone tunnel. The consistent prediction of lowered stress levels at the postero-lateral tunnel wall may lead to bone resorption (according to Wolff's Law), eventually contributing to tunnel enlargement. As such, bone tunnels should be placed as far away from the central cancellous bone as clinically possible. Caution must be used because if a tibial tunnel is placed too anteriorly, knee extension is limited because the graft impinges on the roof of the intercondylar notch (Fu et al., 1999). In addition, an anteriorly placed tunnel increases the chance of penetrating the cortical bone, which will cause large stress level increases in the cortical bone itself.

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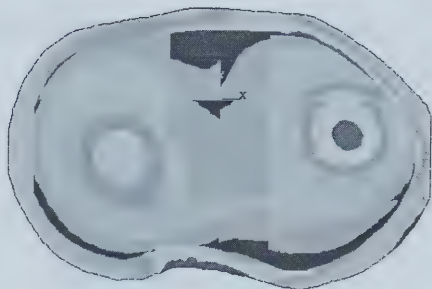
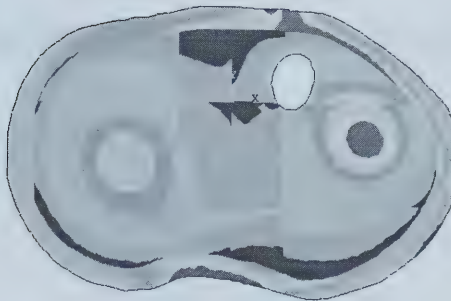
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With Bone Tunnel

Without Bone Tunnel



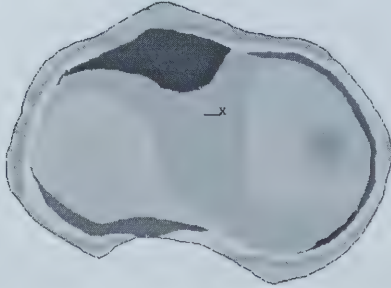
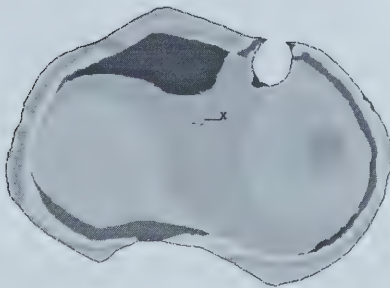
(a)



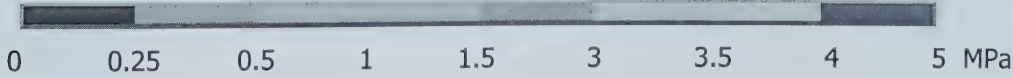
(b)

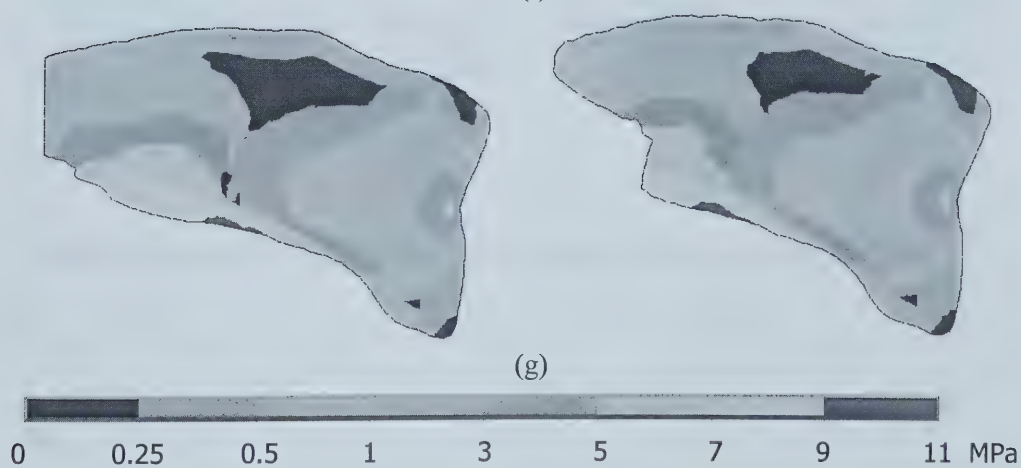
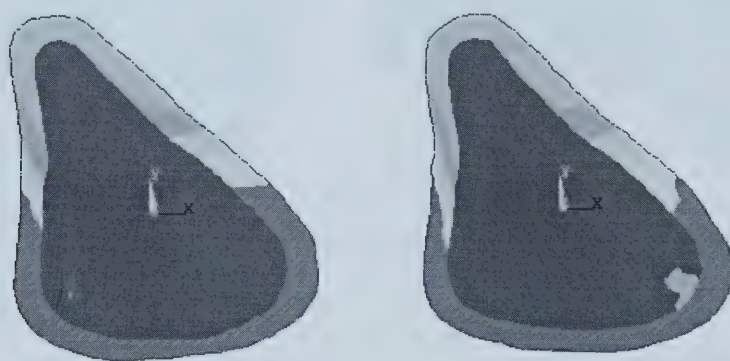
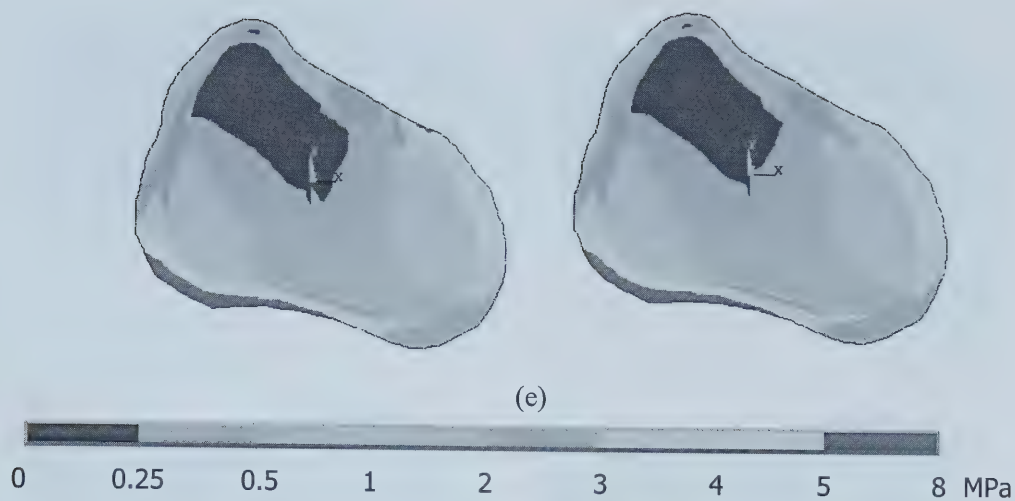


(c)



(d)





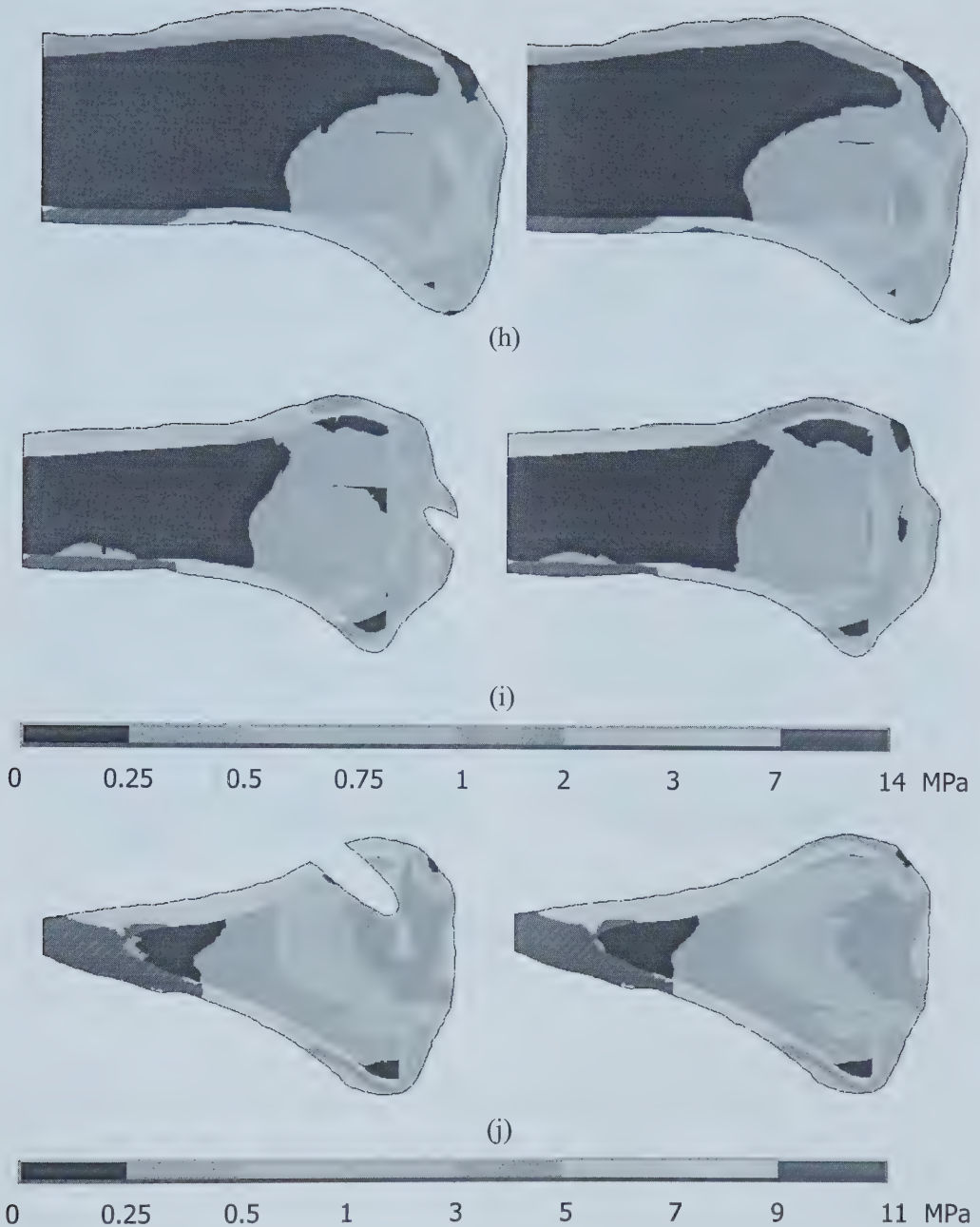


Figure 8.3: Transverse and sagittal sections of tibia FE models comparing the von Mises stresses predicted in a model with a bone tunnel and without a bone tunnel. Bone tunnel was placed at 25 degrees anterior to the midcoronal plane and 10 degrees medial to the midsagittal plane. Transverse sections are taken at (a) 15 mm, (b) 20 mm, (c) 25 mm, (d) 30 mm, (e) 40 mm, and (f) 60 mm below the joint line. Sagittal sections are taken at (g) 15 mm lateral, (h) 5 mm lateral, (i) 5 mm medial, and (j) 15 mm lateral from the midsagittal plane.

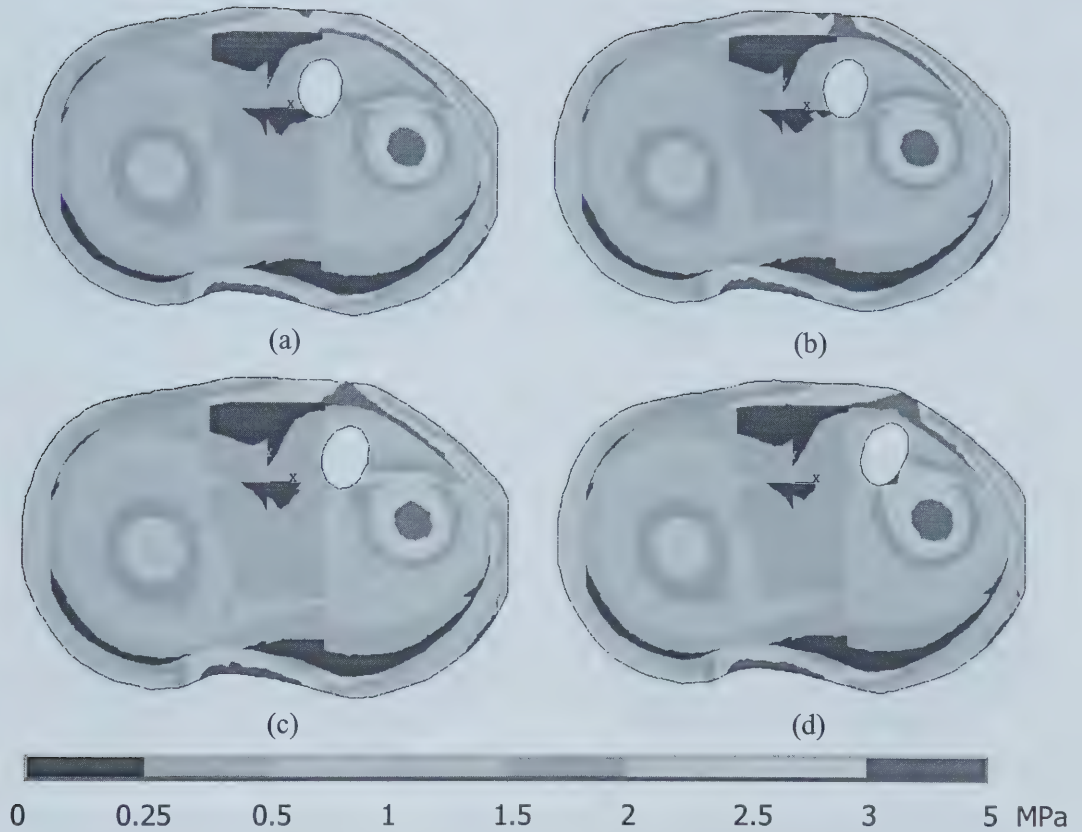


Figure 8.4: Von Mises stress contour plots of transverse sections of tibia FE model at 20 mm below joint line comparing tunnel rotation about the sagittal plane. The tunnel is rotated more medially as sections progress from left to right: (a) 10 degrees, (b) 15 degrees, (c) 20 degrees, (d) 27 degrees from midsagittal plane. The tunnel is rotated 35 degrees anteriorly from the midcoronal plane.

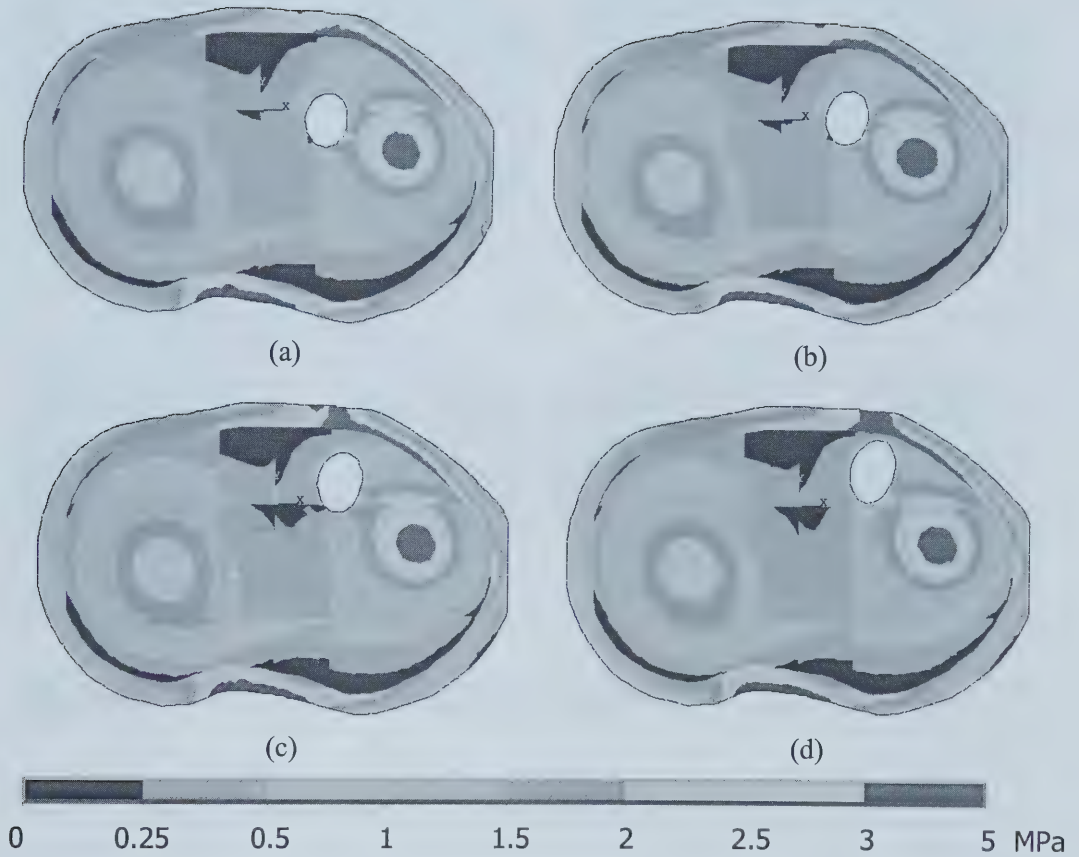


Figure 8.5: Von Mises stress contour plots of transverse sections of tibia FE model at 20 mm below joint line comparing tunnel rotation about the coronal plane. The tunnel is rotated more anteriorly as sections progress from left to right: (a) 25 degrees, (b) 30 degrees, (c) 35 degrees, (d) 40 degrees from midcoronal plane. The tunnel is rotated 15 degrees medially from the midsagittal plane.

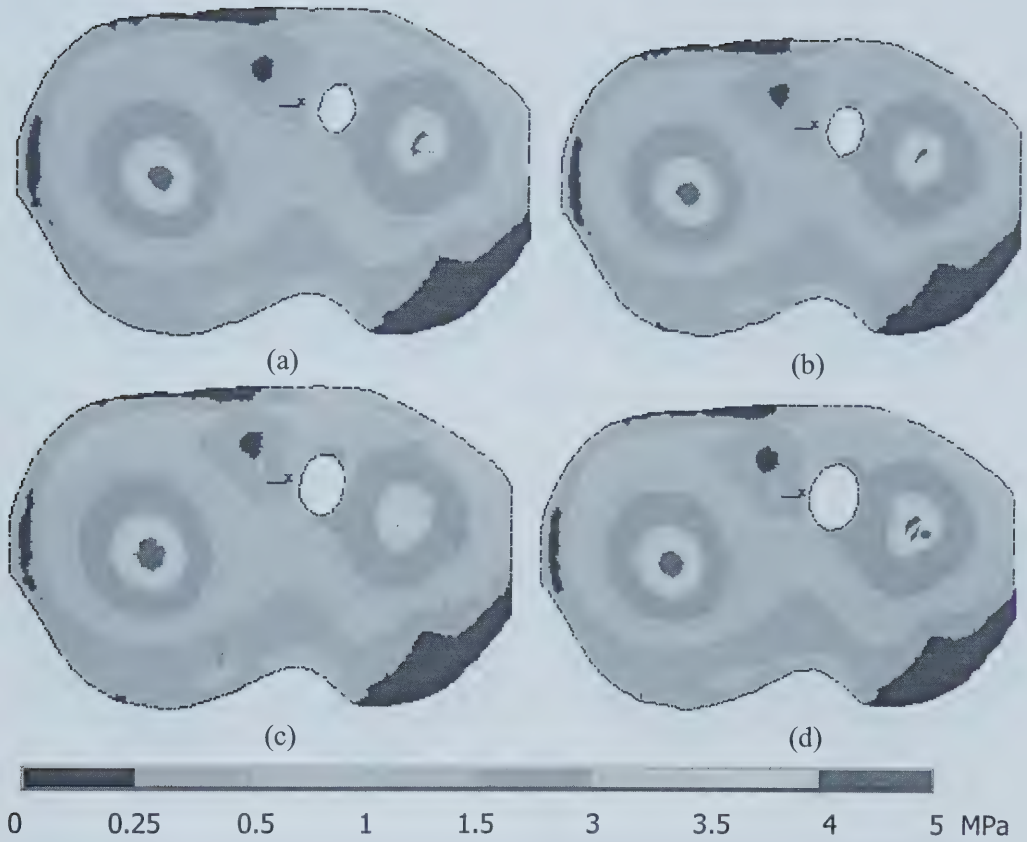


Figure 8.6: Von Mises stress contour plots of transverse sections of tibia FE model at 15 mm below joint line comparing tunnel diameters. The tunnel diameter is increased as sections progress from left to right: (a) 6 mm, (b) 7 mm, (c) 8 mm, (d) 9 mm. The tunnel is rotated 15 degrees medially from midsagittal plane and 35 degrees anteriorly from midcoronal plane.

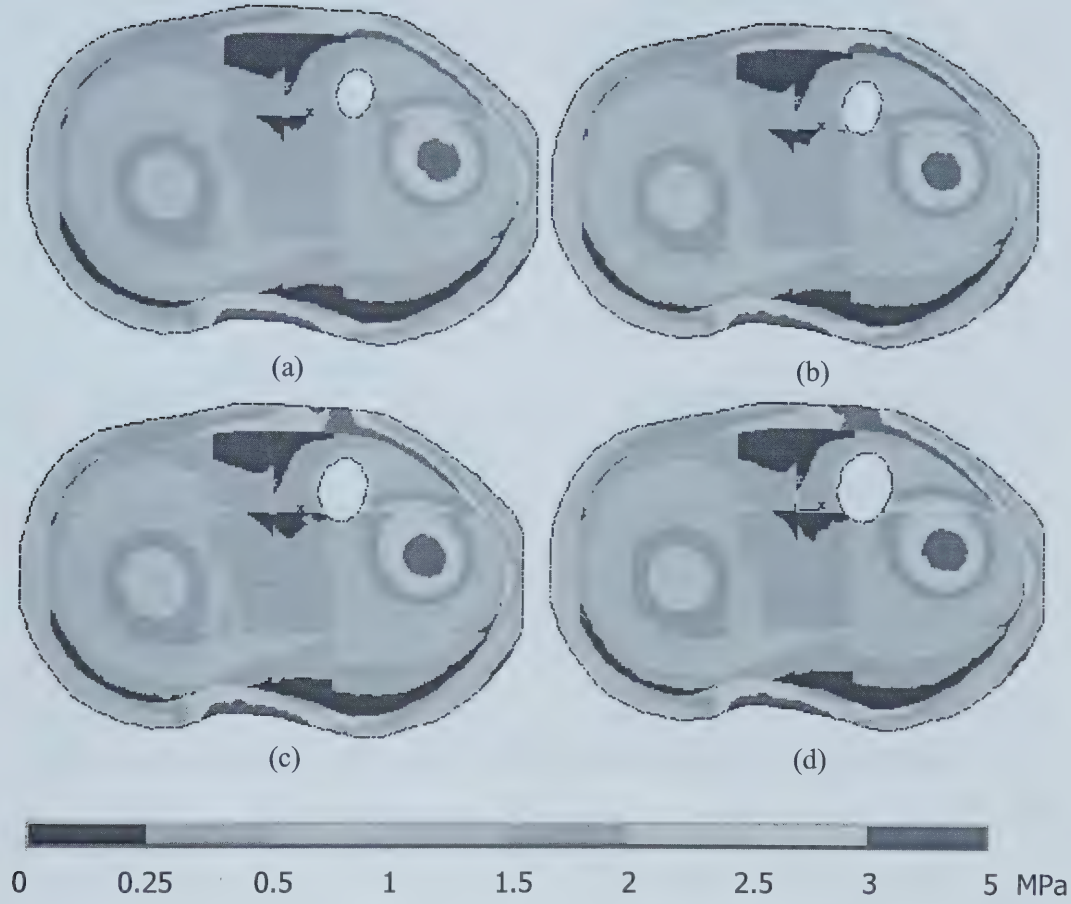


Figure 8.7: Von Mises stress contour plots of transverse sections of tibia FE model at 20 mm below joint line comparing tunnel diameters. The tunnel diameter is increased as sections progress from left to right: (a) 6 mm, (b) 7 mm, (c) 8 mm, (d) 9 mm. The tunnel is rotated 15 degrees medially from midsagittal plane and 35 degrees anteriorly from midcoronal plane.

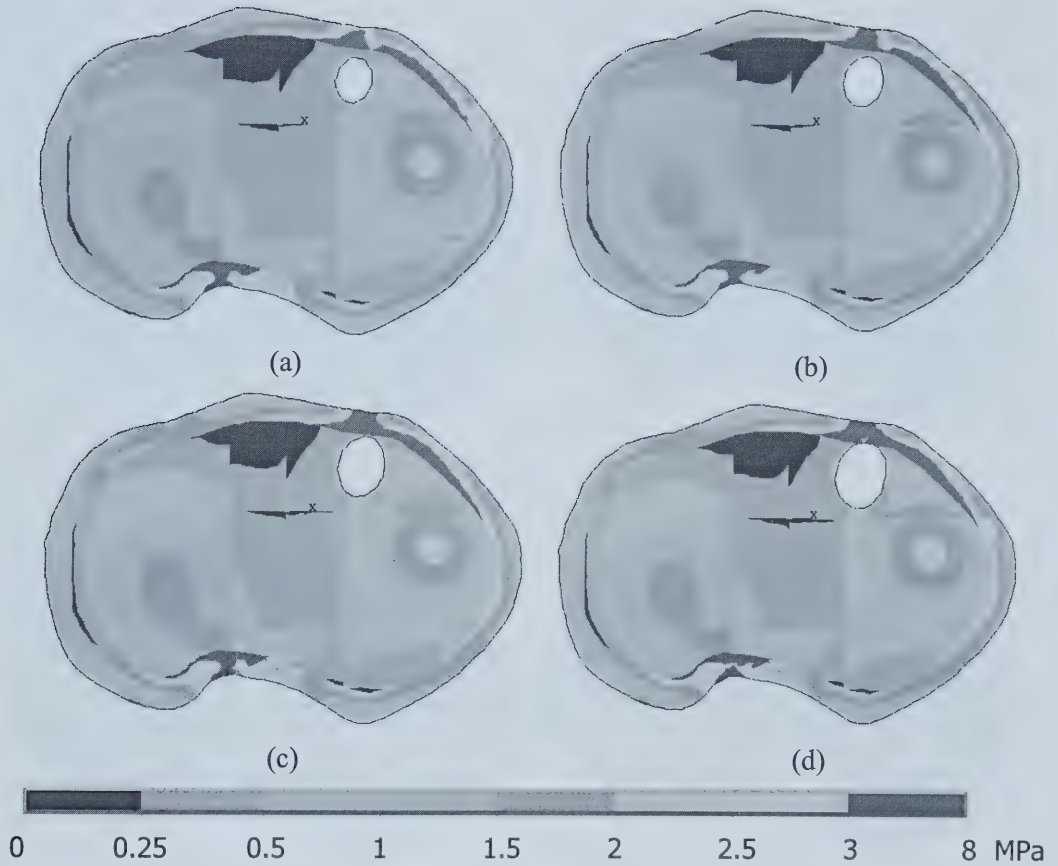


Figure 8.8: Von Mises stress contour plots of transverse sections of tibia FE model at 25 mm below joint line comparing tunnel diameters. The tunnel diameter is increased as sections progress from left to right: (a) 6 mm, (b) 7 mm, (c) 8 mm, (d) 9 mm. The tunnel is rotated 15 degrees medially from midsagittal plane and 35 degrees anteriorly from midcoronal plane.

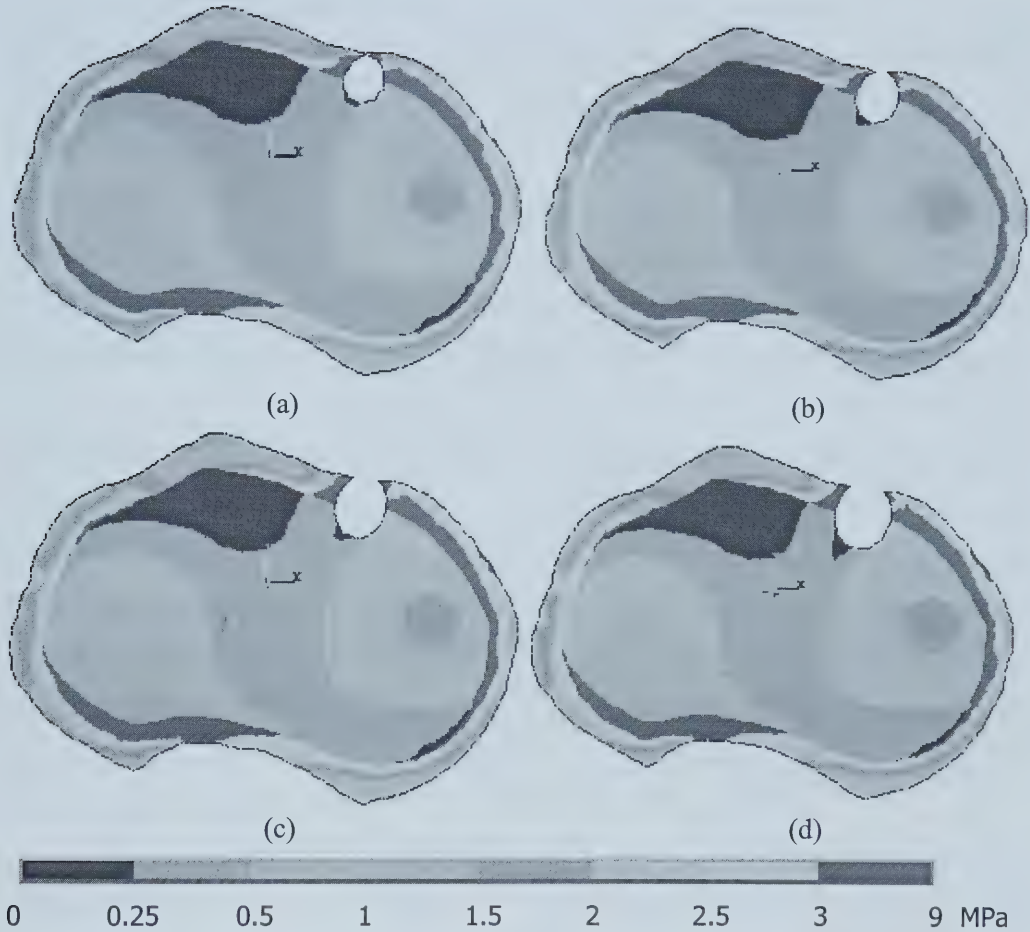


Figure 8.9: Von Mises stress contour plots of transverse sections of tibia FE model at 30 mm below joint line comparing tunnel diameters. The tunnel diameter is increased as sections progress from left to right: (a) 6 mm, (b) 7 mm, (c) 8 mm, (d) 9 mm. The tunnel is rotated 15 degrees medially from midsagittal plane and 35 degrees anteriorly from midcoronal plane.

Although no FE studies have investigated the stresses around an isolated bone tunnel, the computational results of this study may be compared with observations in several clinical studies. Clinical investigations have found bone tunnel enlargement in the posterior portion of the tibial tunnel (Nebelung et al., 1998; Segawa et al., 2003). In a poorly placed tunnel, a graft (fixated with a button) may shift to a more physiologic position, causing the occurrence of localized osteoresorption at the tunnel wall. A greater number of tunnel enlargements were seen in tibial tunnels placed more anteriorly (Segawa et al., 2001). In our models, a greater exposure to higher stresses in tibial tunnels rotated more anteriorly, with the higher stresses located in the postero-medial tunnel wall, was seen. In addition, stress-shielding consistently occurred in the postero-lateral tunnel wall. Bone tunnel enlargement was also seen to increase towards the intra-articular cavity. The stresses were also predicted to be larger in the more proximal bone surrounding the tunnel. While the windshield wiper effect (i.e. the anterior-posterior graft motion within the bone tunnel) is likely the main contributor to the tunnel enlargement (Segawa et al., 2001; Hoher et al., 1998), the internal stresses on the tunnel aside from the graft itself may add to the enlargement. Our FE simulations did not include the effects of the graft on the tunnel, nevertheless, a major advantage of the model simulations was their ability to isolate the regions of stress shielding in the tunnel and, as discussed above, provide guiding predictions for bone resorption patterns that can lead to tunnel widening.

The computational finite element tool will shortly be applied to the investigation of tunnel placement in the femur for ACL reconstruction. A similar methodology to the one used in tibial tunnel placement will be applied.

8.1.2 Cortical Stress from EndoButton

A femur FE model with two concentric tunnels was created to represent the operative procedure to the femur during ACL reconstruction surgery (Fig 8.10). The smaller guide tunnel was 4mm in diameter; the larger tunnel was 8.5mm in diameter. Anisotropic and inhomogeneous bone properties were assigned to the model. A rigid

boundary condition was placed at the proximal base of the femur. The button was assumed to have a rectangular shape with two holes for looping of the polyester tape, and was modeled to contact the femoral cortex at 4 surface points. It was also assumed to be pulled on by a 200 N force directed along the tunnel axis. The 200 N force is an approximation of the graft tension at full extension of the knee during gait (Harrington, 1976; Pandey and Shelburne, 1997). The resultant forces on the 4 points were calculated using force and moment equations knowing the button geometry, and then assigned to the model. A force-balance equation was used to ensure that the four reaction forces were equivalent to 200 N. Three moment equations were used to constrain rotation in all three directions.

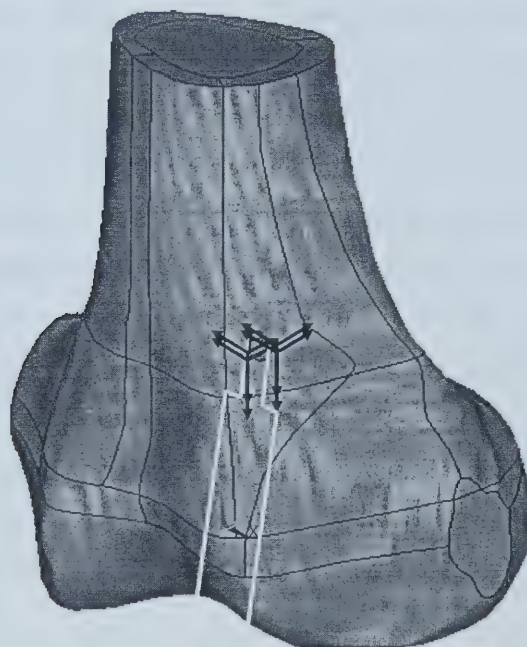


Figure 8.10: A schematic of the tunnel and loading represented in the femur FE model. The solid white line depicts a 4 mm diameter guide tunnel distally enlarged to an 8.5 mm diameter. The arrows represent loads placed at the four contact points between the button and the cortex.

As buttons have been recommended for use in people over 45 years of age, it could potentially be implemented in patients with weak bone stock. A strength of this FE tool is the ability to easily incorporate different sets of material properties. A comparison of cortical bone stresses was made between femur FE models with 45-year-old and 65-

year-old bone. The cortical properties for the 45- and 65-year-old bones were determined by extrapolating the stiffness and shear modulus (using the empirical relationships of Burstein et al. (1976)) from experimental data of 60-year-old bone (Rho, 1992). The cortical and cancellous properties for the 45- and 65-year-old bones were determined by extrapolating the stiffness and shear modulus (as described in Section 7.4) from experimental data of 60-year-old bone (Rho, 1992). This is the first finite element study to investigate the effects of internal stresses in bone with respect to aging from a macroscopic point of view. A finer 1 mm mesh was applied at the lateral cortex at the aperture of the bone tunnel to provide additional stress data in this region. A non-uniform distributed load was applied to the medial and lateral femoral condyles (see Section 6.2) in addition to the button load. Identical loading and boundary conditions were implemented for the two models representing 45- and 65-year-old bone.

The stresses at the cortical bone surrounding the bone tunnel were found to be the largest. It can be seen that even with a 200 N load, a typical load placed on the ACL during walking (Harrington, 1976), the stress on the cortical bone was predicted to be as high as 100 MPa. Stresses greater than 10 MPa were observed around the distal tunnel edge on the cortical surface. Both the stress levels and the stress pattern were nearly identical for the 45- and 65-year-old models (Fig 8.11). Although the mechanical properties of the cortical bone were different for the two FE models, this age difference appears minimal relative to the stresses from the button-like fixation. Therefore, the mechanical effects of the button-type fixation on the cortical bone are not predicted to change greatly for bone between 45 and 65 years of age.

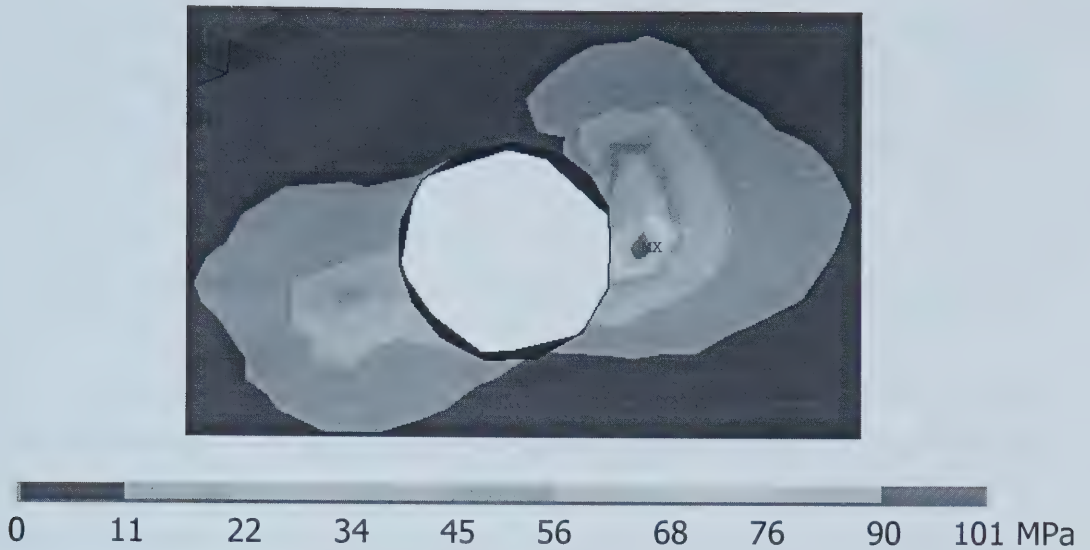


Figure 8.11: Contour plot of von Mises stresses surrounding the cortical bone tunnel surface due to loading from the button. The plot is of a plane view of the stress field, representative of the stresses seen in the models simulating 45- and 65-year-old bones.

Although simplified, the information from this simulation shows that loads produced by the button-type fixation on the cortical surface may be quite high even in normal gait. Activities such as running are likely to increase the stresses on the cortical surface caused by a button. With repeated stressing, microdamages can develop and accumulate in the cortex (Turner et al. 2001) and potentially lead to failure of the bone. The fatigue strength of cortical bone has been tested by numerous investigators (Carter et al., 1981; Carter and Hayes, 1976, 1977b; Gray and Korbacher, 1974; Swanson et al., 1971; Lafferty and Raju, 1979; Choi and Goldstein, 1992), with variations in predicted fatigue strength. The fatigue strength of cortical bone under compression was reported to be approximately 100 MPa for 10^6 cycles. It is estimated that approximately 10^7 loading cycles are applied to bone tissue over a 10 year period (Swanson et al., 1971). Our models predicted a maximum stress of 100 MPa on the cortex assuming full extension during gait (for the given button geometry). However, the cortical stress is expected to increase for more physical activities, and may possibly increase to 200 MPa. While there have been no documented cases of cortical fracture originating from loads produced by button-type fixation, repeated compression of the cortex over the course of a 10 year period may fracture the bone underneath the button and compromise the reconstruction

for recipients of the treatment over the age of 45 (note that we have not found a long term (> 3 yrs) clinical study). This conclusion is strictly based on the assumption that bone remodeling is not present. With bone remodeling, the fatigue life of the cortex may be extended.

8.2 Application of Finite Element Tool to Total Knee Replacement

Knee replacement (or arthroplasty) is a surgical procedure that decreases pain and improves the quality of life for people with severely arthritic knees. Arthritis refers to damage of the articular cartilage and typically arises from wear and tear of the joint (osteoarthritis) or the immune system mistakenly attacking the joint (rheumatoid arthritis). Depending on the severity of the arthritis, there are two types of knee replacements: total knee replacements or unicompartmental (or partial) knee replacements. In total knee replacement (TKR) surgery, the damaged bone is cut away and replaced with a prosthesis. The prosthesis prevents the bones from rubbing together, providing a smooth knee joint. An example of a total knee replacement prosthesis is shown in Fig 8.12.



Figure 8.12: A total knee replacement prosthesis. Reproduced from <http://www.orthopaedicclinic.com/tkrgeneral.html>

The tibial prosthesis component consists of a metal tray attached directly to the bone and a plastic spacer that provides the bearing surface. The metal tray is secured in the tibia with differing methods depending on the type of prosthesis. Often, a metal post is attached to the tray and is cemented to the tibial cancellous bone. However, newer prostheses contain pegs attached to the tray which can be press-fit into the bone and do not require the use of bone cement. Other cementless methods involve fixating the tray to the bone using screws. Revision knee replacement surgery is often associated with aseptic loosening of the tibial prosthesis component rather than the femoral component. Therefore, in this study, the finite element tool is applied to the design of tibial prostheses.

8.2.1 Tibial Prosthesis Design

A demonstration of the FE simulation tool is presented with a tibia knee prosthesis FE model. A major clinical problem with knee prostheses secured with polymethylmethacrylate (PMMA) bone cement is loosening of the tibial component. The loosening may be due to failure of the PMMA-bone interface, PMMA fracture, bone fracture, or bone resorption (as predicted by Wolff's Law). FE models have been used by a number of investigators to study the stresses related to the tibial knee prosthesis. Askew and Lewis (1981) developed a 2D model containing anisotropic and inhomogeneous bone. 3D asymmetric FE models used to investigate tibial knee prostheses include studies by Lewis et al. (1982), Little et al. (1986), and Hashemi and Shirazi-Adl (2000). These FE models contained heterogeneous bone, but did not take into account the anisotropy of the bone. Our comparison of anisotropic and isotropic FE models found that the anisotropy of the bone must be taken into account or predictions may be inaccurate. By taking into account both the anisotropic and inhomogeneous aspects of bone, the tool presented here can provide valuable mechano-biological information of the stresses surrounding a loaded tibial knee prosthesis. Different types of knee prostheses can be combined with the FE model and their stress states compared.

The tibial knee prosthesis was created and assembled to the tibia in Pro/Engineer® (Fig 8.13) and exported to ANSYS®. A typical metal prosthesis component consists of a

plate on a post fixed in the tibial model by a surrounding layer of PMMA bone cement (Fig 8.14). A layer of ultra-high molecular weight polyethylene (UHMWPE) was modeled atop the tibial tray. Four FE tibia prosthesis models were created incorporating a straight circular post, a tapered circular post, a straight elliptical post, and a straight circular post and pegs (Fig 8.15). The metal was assigned a Young's modulus of 200 GPa and a Poisson's ratio of 0.3 (Murase et al., 1983). The bone cement was given a Young's modulus of 2 GPa and a Poisson's ratio of 0.23 (Murase et al., 1983). The UHMWPE was assigned a Young's modulus of 1083 MPa and a Poisson's ratio of 0.45 (Hashemi and Shirazi-Adl, 2000). In each model, a 2000N distributed loading was placed onto the polyethylene surface, and the base of the tibia was fully constrained. While the polyethylene surface is noticeably different from the intact tibial joint surface, the same non-uniform distributed loading was used for both TKR and non-TKR FE models in this study. However, it is an improvement upon the loading conditions used in other FE studies involving TKR models, and provides a more realistic depiction of the stress state. It is acknowledged that this is not the most physiologically realistic loading condition, but it is indeed sufficient for a parametric analysis of the TKR designs.

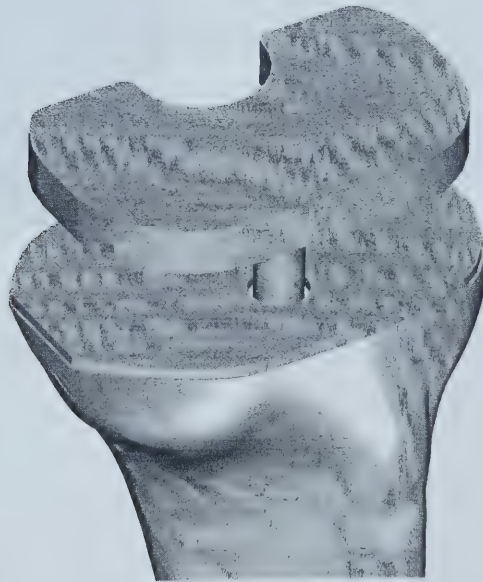


Figure 8.13: A tibia prosthesis created and assembled in Pro/Engineer[®]. A tray, straight circular post, PMMA, and tibia are shown here. This solid model is exported to ANSYS[®] for stress analysis.

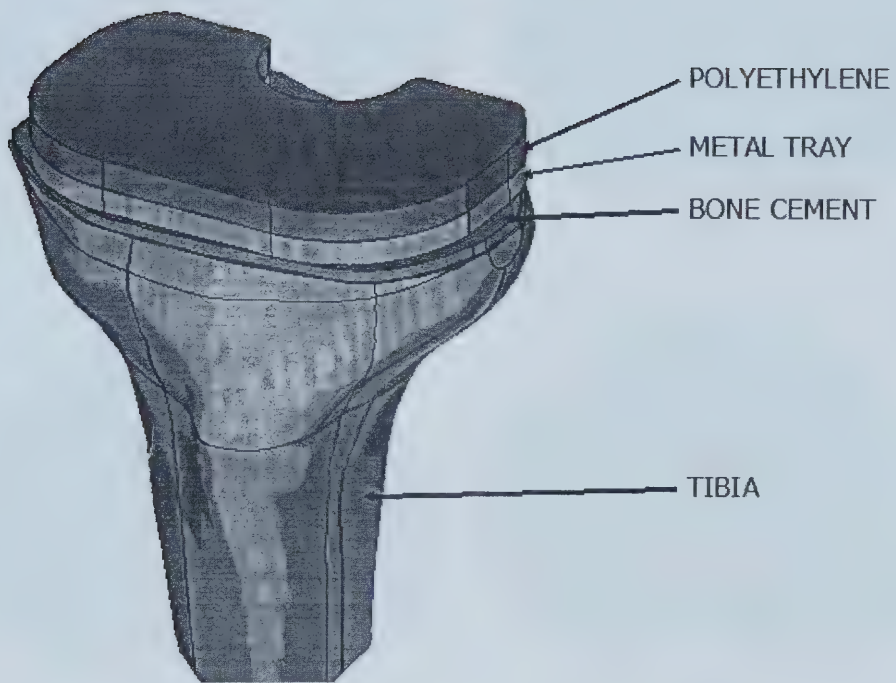


Figure 8.14: FE model containing tibia prosthesis. A layer of bone cement was modeled in addition to the metal and polyethylene prosthesis to mimic the clinical practice.

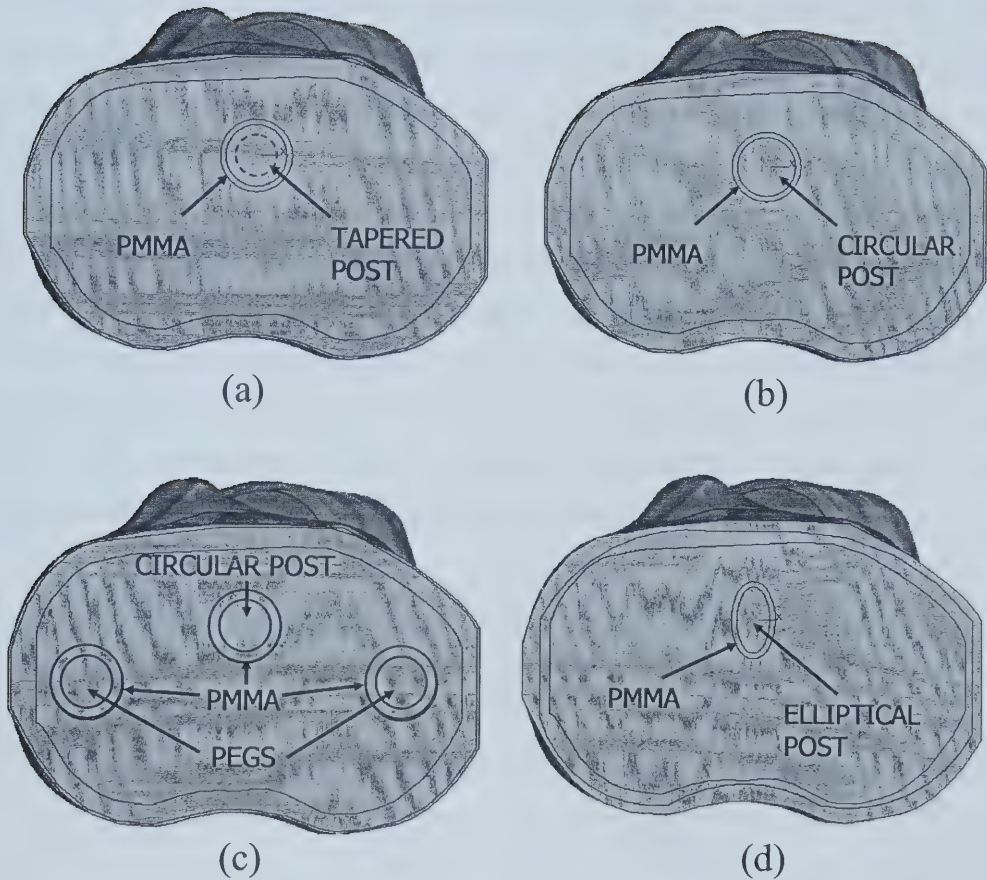


Figure 8.15: Tibia FE prosthesis model with (a) tapered post, (b) circular post, (c) circular post with pegs, and (d) elliptical post. All FE models contain posts and pegs attached to bone with polymethylmethacrylate (PMMA) bone cement.

The distribution of the total load from the prosthesis into the cancellous bone can be determined by examining the cement-bone interface. Forces on the bone were examined for the models without pegs at three locations: beneath the cement layer, around the periphery of the cement which surrounded the metal post, and beneath the cement at the bottom of the post. The load fractions for the three types of posts are

shown in Table 8.1. The bone cement interface area was approximately the same for all three models. The area fractions for the three types of post are shown in Table 8.2. In all three models, the majority of the load is distributed to the bone beneath the plate. The plate compression was almost identical regardless of post shape. However, the distribution of forces around the post varied depending on the shape of the post. Tapering the post produced a slightly greater surface area compared to the straight circular post. However, increasing the surface area did not increase the proportion of the total load carried by the post. Instead, it had the opposite effect by reducing the amount of shear force at the post by 2%. By using an elliptically shaped post with approximately the same post surface area, the amount of shear force increased by 4% compared to the straight circular post. By tapering the straight circular post, the surface area at the base of the post was approximately reduced by half. However, instead of reducing the amount of compression beneath the post, an increase in the proportion of post base compression was seen. While sharing approximately the same post base fractional area, the elliptical post had less compression at the cement-bone interface than the straight circular post.

Table 8.1: The load fractions calculated at the cement-bone interface. The load was assumed to be divided among the three locations: beneath the plate, beneath the post, and surrounding the shaft. Three different post shapes were investigated: a straight circular post, a tapered circular post, and a straight elliptical post.

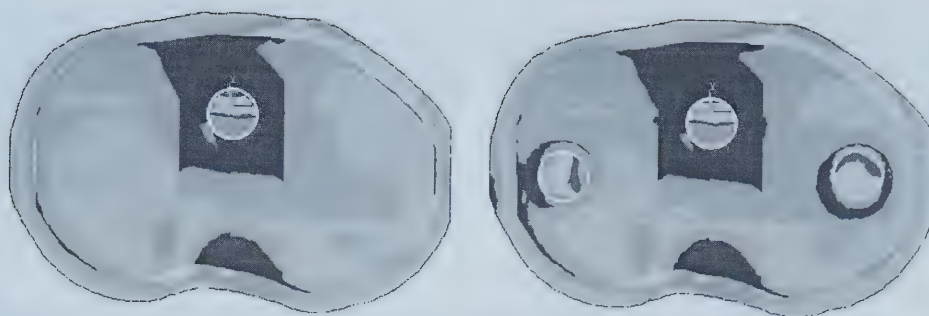
Post shape	Plate compression	Post base compression	Post shear
Straight circular	0.88	0.08	0.04
Tapered circular	0.89	0.09	0.02
Straight elliptical	0.89	0.03	0.08

Table 8.2: The area fractions at the cement-bone interface. The areas beneath the plate, beneath the post, and surrounding the shaft were determined. The contributions of each separate area were calculated as a fraction of the total interface area. The surface areas were used to determine the load fractions in Table 8.1.

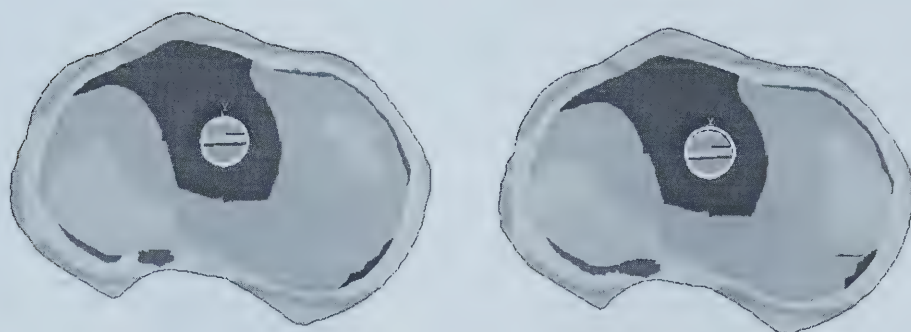
Post shape	Plate	Base	Shaft
Straight circular	0.69	0.02	0.29
Tapered circular	0.68	0.01	0.31
Straight elliptical	0.66	0.02	0.32

As the majority of the load is carried in the bone beneath the plate, changing the post shape may not change the force beneath the plate. A high level of stress is expected to occur at the base of the prosthesis post. If the surface area of the post base can be reduced concurrently with the amount of post compression, the stress level can be reduced. Tapering a post increases the surface area of the shaft and reduces the surface area of the base. Therefore, an increased shear force and reduced post compression is expected with tapering. However, this effect was not observed. Instead of increasing shear force, tapering the post increased compression beneath it. The increase of compression beneath the post coupled with the decrease in the area beneath the post greatly increased the stress. Therefore, tapering the post did not have the intended effect of reducing the load fraction beneath the post. This effect was seen with the introduction of an elliptical post. While maintaining approximately the same interface area as the tapered and straight post models, using an elliptical post can relieve bone compression by redistributing the force from the base of the post to its periphery. Such a design may benefit clinically as a more even stress distribution will result in a lower rate of bone resorption.

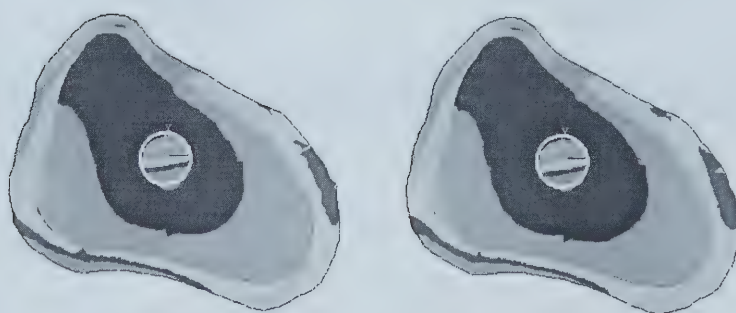
Pegs may be placed in prostheses to increase the contact area between the prosthesis and bone, as well as to prevent rotation of the prosthesis. The insertion of the pegs into the cancellous bone may cause an increase in the internal stresses. This tool was used to investigate the effects of adding metal pegs in tibial prostheses. Two FE models described above were used to analyze the stresses: the model with a straight metal post, and the one with two metal pegs in addition to a post. All prostheses were fixed to the tibia with a surrounding layer of PMMA. Material properties, loading and boundary conditions were identical to those described above.



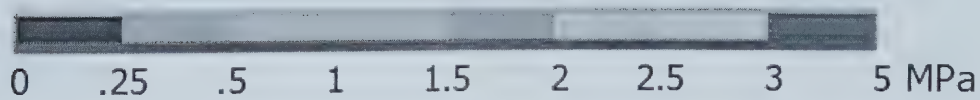
(a)

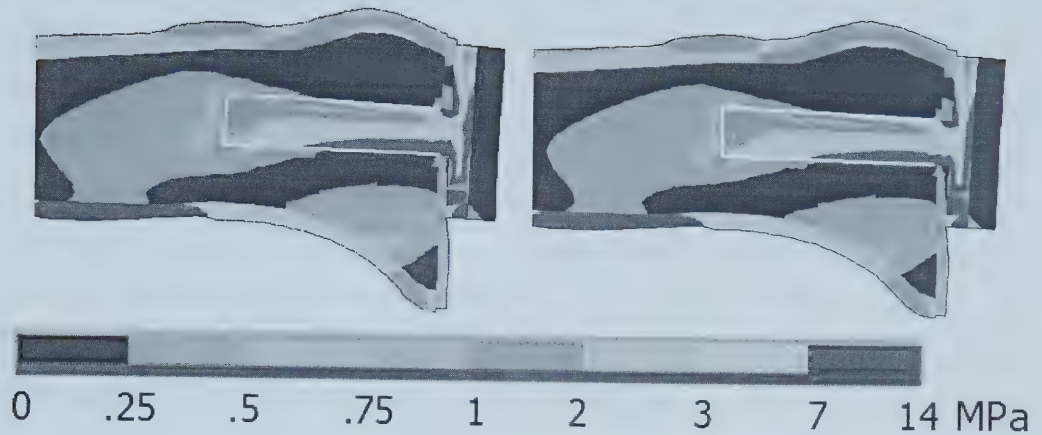


(b)

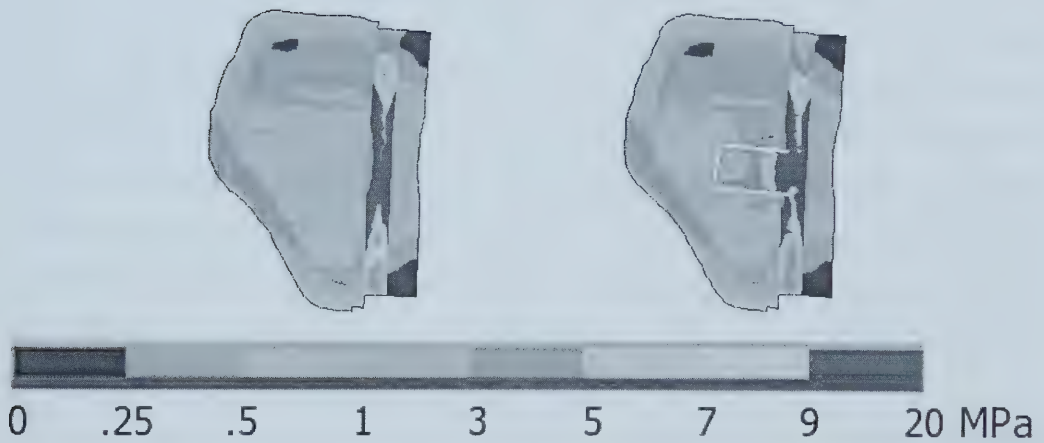


(c)





(d)



(e)

Figure 8.16: Von Mises stress contour plots of transverse sections of tibia FE model at (a) 20 mm, (b) 30 mm, (c) 40 mm below the joint line. Sagittal sections at (d) 0mm, (e) 25mm lateral from the midline are also shown. The left-hand column shows plots of the model containing a metal post; the right-hand column shows plots of the model containing a metal post and two pegs. White circles/boundaries depict locations of the post and pegs.

Sections along the length of the tibia models were compared using von Mises stress contour plots (Fig 8.16). The addition of pegs caused a reduction of stress in the surrounding bone in the periphery (Figs 8.16a and e). In the more proximal bone (Figs 8.16a and b), the maximum stress (located in the inner cortex) was lower in the model containing pegs. While lowering stress in the proximal bone, the pegs created an increase in stress around their base (Fig 8.16e). Close proximity of the lateral peg to the cortical shell produced an increase in the cancellous bone stress at the base of the peg (Fig 8.16e). The addition of pegs did not influence the stresses around the long metal post as Fig 8.16d showed the contour stresses to be nearly identical.

In addition to preventing rotation of the prosthesis, the pegs reduced the stresses in its periphery without increasing the stress in the post. Lewis et al. (1982) found that the presence of posts reduces the cement-bone interface normal stresses beneath the plate. The use of multiple posts/pegs can enhance this outcome because of the increased surface contact area between prosthesis and bone. The partial transmission of the applied compression load via shear forces across the vertical interfaces between bone and pegs (Hashemi and Shirazi-Adl, 2000) is central to this occurrence. Another advantage of using a prosthesis containing both post and pegs is reducing the tilting of the prosthesis. When using a single post, the post appears to act as a fulcrum, resulting in increased tilting of the prosthesis (Lewis et al., 1982). However, the addition of pegs can remove this fulcrum and prevent tilting of the prosthesis while enhancing the stress field distribution in the bone.

The reduction of bone stress around the prosthesis may lead to resorption of bone mass due to the prosthesis adopting part of the mechanical bone stresses (Huiskes, 1998). This form of aseptic loosening may cause a cavity to form around the prosthesis resulting in painful joints. A comparison of the cylindrical and elliptical post FE models predicted a greater stress-shielding of the bone around the cylindrical post. Therefore, varying the post shape of a tibial prosthesis may increase mechanical stress at the bone-prosthetic interface. The stressing of the bone is expected to reduce bone resorption in accordance with Wolff's Law. As discussed above, cylindrically-shaped pegs were seen to reduce stress in the proximal tibia. Bone resorption may possibly occur in the region surrounding these pegs, which may be reduced by varying the shape of the pegs.

A clinically beneficial application of this finite element tool is mechanical analysis of tibial prosthesis design. Among the many knee prosthesis designs (in 1997, 37 different prosthesis designs were being marketed in the United Kingdom (Liow and Murray, 1997)) one of the most common failure mechanisms of total knee arthroplasties is component loosening in both cemented and uncemented prostheses (Sharkey et al., 2002). Shear stress is a central mechanical cause of implant loosening (Walker et al., 1981). The tool presented in this study can be used to model any number of designs and analyzed for potential mechanical failure mechanisms. The realistic stress distributions generated by considering anisotropic and heterogeneous bone properties allowed for a more comprehensive analysis of the influence of post design. A comparison of straight, tapered, and elliptically shaped post prostheses revealed that the effects of post design may not always be intuitive (e.g. tapering a post reduces post shear and increases post compression, while an elliptical design generates the opposite effect). The addition of pegs to a central fixation post design prevents tilt and rotation of the tibial prosthesis but may induce bone resorption due to reduced bone stresses. No particular model of prosthesis was modeled in this investigation. However, critical analyses of poorly designed prostheses using the developed computational finite element tool may assist in reducing component failure, adding to the clinical value of this tool.

Chapter 9: Summary, Conclusions and Future Work

9.1 Summary and Conclusions

Potential failures in knee reconstructions and total knee replacements can be detected with accurate stress analyses of the internal stresses within the knee. Many studies have adopted the finite element method to assist in the development of knee replacement implants. However, assumptions have been made in the development of previous finite element models of the knee which have compromised the accuracy and applicability of these models. A new simulation tool is described in this thesis which provides a comprehensive approach to the stress analysis of the femur and tibia.

In Chapter 3, the three-dimensional geometries of the femur and the tibia were introduced. The geometries are based on anatomically realistic surface geometries of third-generation composite bone models. The accurate geometries were generated using a laser scanner and rendered using a powerful CAD package, Pro/Engineer®, and are available at the International Society of Biomechanics website.

Chapter 4 fully described the methods used to assign material properties to the FE models. Bone is a complex biological material and cannot be overly simplified in FE models without compromising their accuracy. Cortical and cancellous bones exhibit both directionally-varying (anisotropy) and spatially-varying (inhomogeneity) material properties. The complex nature of these bone types were considered in the finite element models developed in this study. The bone in the femur and the tibia FE models were divided up into numerous volumes to allow for the accommodation of orthotropic and inhomogeneous cortical and cancellous bone properties. The assigned properties were based on experimental data reported in literature. The ease of bone property incorporation enables this model to be used for studies of aging in relation to internal stresses of the femur and the tibia.

A physiologically representative loading scheme was described in Chapter 6. A general method is shown which can be used to incorporate experimentally determined load patterns. As an example, a non-uniform distributed loading pattern available in

literature was incorporated into the FE models. With a sufficiently accurate description, any load pattern can be incorporated into these models.

In Chapter 7, internal and external validations were performed on the FE models. Local and global accuracy checks were used to determine the quality of the numerical solutions. Stress and displacement convergence analyses confirmed good convergence of solutions for the femur and tibia FE models. A strain energy error convergence further confirmed the quality of the solutions.

The developed tibia FE model was compared with *in vitro* strain studies performed on cadavers. The FE models demonstrated a good correlation with two experimental studies. The same trends of strain values shown in the experimental studies were also observed in the tibia FE model in this study. This validation showed that the developed FE model has the potential to become a valuable assistant in the discussion of clinical questions (e.g. the evaluation of different types of knee joint prostheses or tunnel placements in ligament reconstruction surgeries).

The incorporation of anisotropic and inhomogeneous bone allows for a more realistic representation of stresses in the bone than previous FE models. Higher stresses were found in the peripheral cancellous bone and the surrounding cortex when incorporating anisotropic and inhomogeneous bone instead of isotropic and homogeneous bone. Incorporating homogeneous bone may give incorrect predictions of highly stressed cancellous bone areas, while incorporating homogeneous bone may give a consistently low prediction of cancellous bone stress levels. The consideration of both anisotropic and inhomogeneous material properties allows a more accurate depiction of stresses at bone-implant interfaces for applications such as prosthesis design.

Two examples of clinical applications of the developed femur and tibia FE models were demonstrated in Chapter 8. The incorporation of bone tunnels in a tibia model allowed for a representation of stress states in a post-operative situation of an anterior cruciate ligament reconstruction. The bone tunnel caused stress-shielding at the postero-lateral region of the tunnel wall, while increased stress occurred at the postero-medial region of the tunnel wall. The tunnel stresses in the more distal regions of the cancellous bone were in general lower than those in the more proximal regions.

Prolonged stress-shielding may possibly cause bone resorption of the posterior tunnel wall leading to tunnel enlargement, possibly compromising the ACL reconstruction.

The use of buttons for peripheral fixation of a soft tissue graft was found to produce a noticeable amount of stress on the cortex around the bone tunnel for patients over 45 years of age. The stress levels and patterns were nearly identical for the 45- and 65-year-old models. Although the immediate stress levels will not fracture the cortex, repeated compression from the button in addition to a greater stress level may cause microdamage in the cortex and eventually lead to fatigue failure.

The FE tool was also used to evaluate different tibial prosthesis designs. Prosthesis models containing a straight circular post, a tapered circular post, a straight elliptical post, and a straight circular post and pegs were compared. A high level of stress was found to occur beneath the implant post. The elliptical post was found to relieve post base compression more effectively than the tapered post. Prosthesis pegs were seen to reduce stress in their periphery which may lead to resorption of bone mass which may potentially cause aseptic loosening of the prosthesis.

9.2 Future Work

An immediate task is to evaluate femoral tunnel placement in ACL reconstruction using this FE tool. Studies have been published regarding femoral tunnel placement on graft laxity (Markolf et al., 2002) and tunnel enlargement (Segawa et al., 2001) but not its stress effect on the surrounding bone. An investigation of these internal stresses may potentially have clinically positive benefits. A second task that should be explored is the application of different loading conditions (e.g. increasing flexion angles) to expand the analysis of stress states for the above clinical applications.

A central aspect of the developed tool is the consideration of anisotropic and heterogeneous material properties. While the most recent and detailed experimental data have been applied to the model, some assumptions were still necessary. As more detailed experimental data for subchondral, cortical, and cancellous bones become available, they should be incorporated into the FE models to improve the representation of the stress states.

A restriction of the possible clinical value of this tool is that the kinematic boundary conditions are simplified. Although a more physiologically representative distribution and contact area of the loading of the bones have been considered, there was no incorporation of the mechanical influence of other muscles and ligaments acting on the knee, e.g. anterior and posterior cruciate ligaments, quadriceps femoris, or the effect of menisci. Representation of these muscles, ligaments, and menisci are necessary to obtain a numerical simulation tool which more closely reflects *in vivo* conditions. Consequently, further work must be done to optimize the developed FE model focusing on the influence of soft tissues acting around the knee joint.

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Appendix A

A.1 Novel Hybrid Fixation for Anterior Cruciate Ligament Reconstruction

This section contains a compilation of experimental work completed during my Master's studies.

A.1.1 INTRODUCTION

The anterior cruciate ligament (ACL) is one of the most commonly damaged ligaments in the knee. It is estimated that 1 in 3000 people in the general population sustain ACL injuries¹. The ACL has been found to have a poor healing response² and non-operative treatment of the ruptured ACL often is associated with joint instability and increased incidence of meniscal injuries. As a result, more than 50,000 ACL reconstructions are performed in the United States annually³.

Graft fixation is the weakest link in anterior cruciate ligament (ACL) reconstructions during the early postoperative period. Fixation methods with poor structural properties of either stiffness or strength may compromise the clinical outcome⁴. Current fixation techniques involve direct fixation of the graft within the osseous tunnel or periosteal fixation away from the joint surfaces. Soft tissue tendon grafts such as the hamstring tendon graft lack the bony ends of the bone-patellar tendon-bone grafts. One approach to fixating soft tissue grafts may be to directly fixate the tendon to the osseous tunnels with interference screws. Additional fixation strength may be gained by periosteal fixation with other types of fixation devices (such as EndoButtons), but an investigation into this is required. Direct fixation in soft tissue tendon grafts has the advantage of fixation near the joint line. This anatomic fixation increases knee stability, graft isometry, and overcomes biomechanical disadvantages such as graft tunnel motion and the windshield-wiper effect⁵.

Two commonly used types of interference screws are metal and biodegradable. In terms of initial fixation strength, the two types of screws show no significant difference when used with soft tissue grafts⁶. It is known that for bone-patellar tendon-bone grafts, it takes approximately 6 weeks for bone incorporation into the osseous

tunnel⁵. An even longer period is required for soft tissue incorporation into the tunnel. The fixation properties of biodegradable interference screws during its degradation period are relatively unknown⁵. Considering the above, using metal screws might be more appropriate for this study.

As an alternative to direct fixation with interference screws, soft tissue tendon grafts can be secured near the tibial and femoral cortex. The femoral-sided EndoButton and tibial-sided screw and washer have traditionally been used to achieve hamstring tendon graft fixation extra-articularly⁷. A study using human quadriceps tendon showed no significant difference in ultimate tensile failure load when using an EndoButton and polyester tape or biodegradable interference screw. However, stiffness was significantly worse when using the EndoButton⁸.

To improve the strength and stiffness of graft fixation, numerous experiments have been performed regarding fixation materials. Studies on the pullout forces and stiffness of soft tissue grafts using biodegradable interference screws^{6,8-11}, titanium interference screws^{6,9-10}, or EndoButtons⁸ have shown that no fixation method can reproduce the ultimate failure strength or stiffness of the native cruciate ligament. Combining different methods can present an alternative to using different materials to increase fixation stress and stiffness. We hypothesized that combining these direct and periosteal methods of fixation would improve the pullout strength and stiffness for soft tissue grafts. Clinically, this combination might also be beneficial for patients with suspected lower bone mineral density, where back-up fixation to the interference screw may be necessary⁵.

To meet the existing need for stronger and stiffer fixation methods to allow current rehabilitation practices, a novel fixation technique has been proposed. To investigate the merit of our hypothesis, an animal model was adopted. The objective of this study was therefore to compare a novel hybrid fixation method (i.e. combining the titanium interference screw with the EndoButton and two Number 5 sutures) with the classical method of only using an interference screw to secure soft tissue grafts. The factors studied were the initial pullout force, stiffness of fixation, and failure modes.

A.1.2 MATERIALS AND METHODS

A.1.2.1 Materials

Porcine patellar tendon was used as soft tissue graft material. Fresh distal porcine femurs were used to model the young human knee (the use of the porcine model is further elaborated in the Discussion section). The patellar tendon and the femur from the same knee were used in each fixation construct.

A round-headed titanium interference screw (RCI, Smith & Nephew Donjoy, Carlsbad, California) was used to secure the porcine patellar tendon against the bone tunnel. A RCI screw of 7 mm thread diameter and 25 mm length was used in a bone tunnel drilled with an 8.5 mm RCI femoral router (RCI, Smith & Nephew Donjoy, Carlsbad, California). An EndoButton (Smith & Nephew Endoscopy, Andover, Massachusetts) with two Number 5 Ethibond sutures (Ethicon, Somerville, New Jersey) secured the soft tissue graft to the femoral cortex.

A.1.2.2 Preparation of Specimens

Sows aged 1.5 to 3 years and weighing 130 to 160 kilograms were sacrificed and their knees were immediately frozen to -20°C . The knees were thawed to room temperature 16 hours prior to testing.

The patella was separated from the knee with the patellar tendon still attached. A 1 cm diameter hole was drilled through the patella so it could be secured to the testing machine. The patellar tendon was released from its insertion on the tibial tubercle. While still attached to the patella, the tendon was sized to a 70 mm length and a width of 14 mm for a snug fit into a 7.5 mm diameter template hole. The tendon was tubularized with two Number 5 Ethibond sutures using a baseball stitch (Fig. A.1). The graft was kept moist with saline until placed into the tunnel.

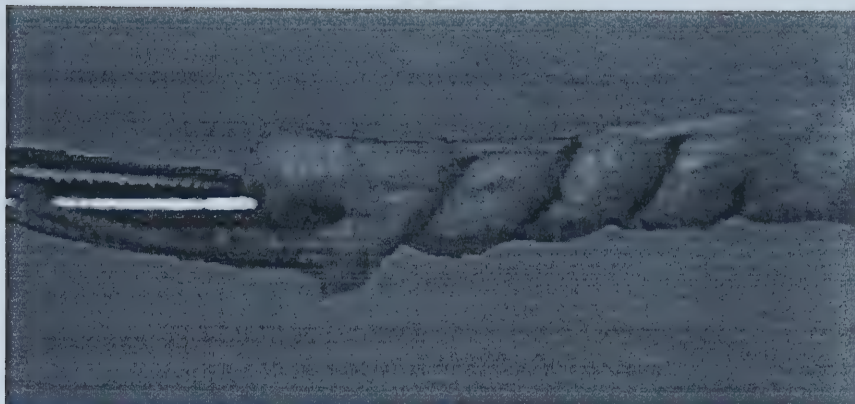


FIGURE A.1: Patellar tendon graft tubularized using a baseball stitch (Number 5 Ethibond suture). The forceps, on the left side, can be seen holding the tendon.

A guide tunnel was drilled offset approximately 1 cm from the intercondylar spine of the femur. This was done for experimental convenience; it is not anticipated that the mechanical testing results would be any different if the tunnel was to be placed in its usual location within the intercondylar notch. The tunnel was drilled parallel to the long axis of the bone using a cannulated guide system. The femoral tunnel was then enlarged to an 8.5 mm diameter and a 40 mm length using a RCI femoral router. The interference screw was inserted using a guide wire and introduced from the proximal side of the tunnel. When periosteal fixation was required, the EndoButton was placed on the cortical surface of the femur seated over the bone tunnel. All four suture strings were anchored to the EndoButton with several knots.

A.1.2.3 Study Groups

The twenty knees were randomly assigned to two groups of 10 knees each. In group 1, a 7 X 25 mm RCI screw was inserted into the interface between the soft tissue graft and the bone tunnel. The screw was seated between the graft and the cancellous surface of the femoral osseous tunnel (Fig. A.2a). In group 2, a 7 X 25 mm RCI screw was inserted in the same manner as in group 1. Also, the Ethibond sutures at the distal end of the graft were tied to an EndoButton seated on the femoral cortex (Fig. A.2b). Considering that porcine bone is relatively dense compared to older human cancellous bone, to eliminate artifacts like laceration of the soft tissue when placing the interference

screw, it was elected to use a 7.5 mm graft in an 8.5 mm bone tunnel. This graft-tunnel complex is not routinely clinically used; however, the data acquired provides qualitative information about the two fixation groups.

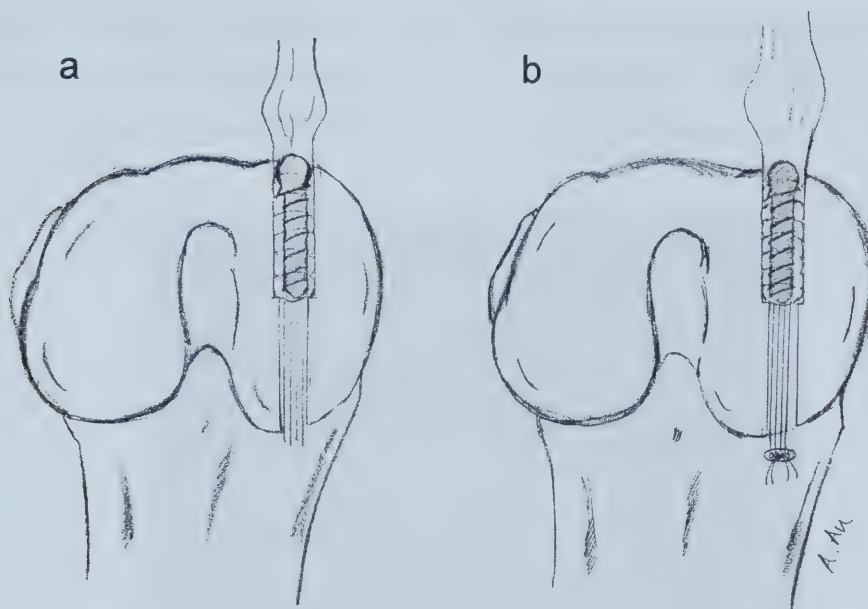


FIGURE A.2: The fixation techniques for the tubularized patellar tendon. (a) fixated with interference screw (note the loose sutures); (b) fixated with interference screw and EndoButton secured with two Number 5 sutures.

A.1.2.4 Testing Procedure

Specimens were mounted onto a tensile testing machine (MTS Model 410.31; Materials Test Systems Corporation, Minneapolis, Minnesota) (Fig. A.3). As shown in Fig. 3, the shaft of the femur was secured to the base of the machine using 4 threaded metal rods with pointed ends biting into the cortex. A smooth rod was passed through the hole in the patella to secure it under the cross arm. The specimens were preloaded with 10 N. The load was applied parallel to the long axis of the bone tunnel until failure to represent a worst-case loading scenario. A rate of 3 mm/second was used to load the specimen. This speed is well within the loading speeds used in the literature (e.g. Caborn et al.⁶ used 0.33 mm/second and Kurosaka et al.¹² used 30 mm/second). Failure mode was determined by visual inspection and recorded as graft sliding past screw,

midsubstance graft failure, or suture tearing through graft. Ultimate tensile load and stiffness were determined from the load-elongation curves. Stiffness was determined by fitting a straight line using least squares method to the rising slope of the load-elongation curve (i.e. slopes A and B in Fig. A.4). The peak of the load-elongation curve was considered to be the ultimate failure load. Computerized data collection was facilitated using a National Instruments DAQ 1200 card (National Instruments, Austin, Texas).

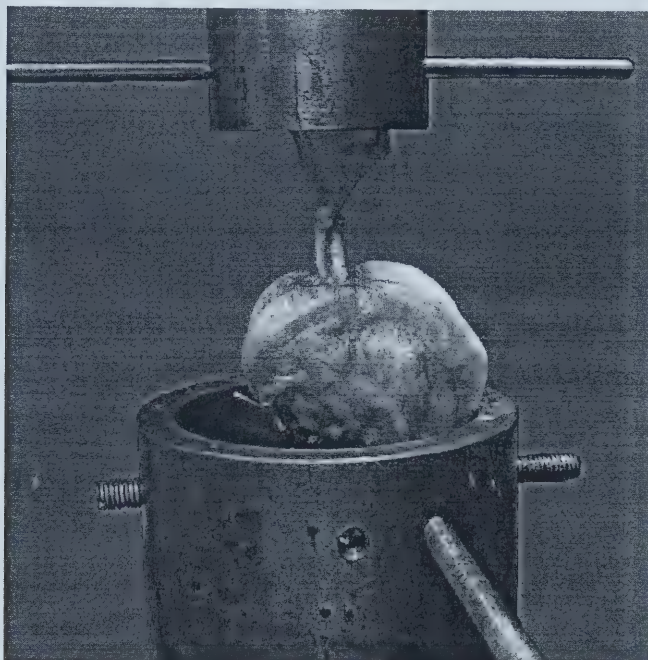


FIGURE A.3: Specimen mounted onto a MTS machine. A metal rod secures the patella to the base of the cross arm (top of the picture) while four threaded metal grips secure the shaft of the porcine femur.

A.1.2.5 Statistical Analysis

After the data were confirmed to have a Gaussian distribution, it was ascertained the number of specimens used were sufficient to warrant statistical analysis of the results for the ultimate failure loads. A power analysis of 87% was found for the ultimate failure loads. For both groups the mean value, the empirical standard deviation, and the confidence limits with a probability of 95% were determined. A one-tailed t-test at the

95% confidence level was used to compare groups 1 and 2. The t-test was used to compare the means of the ultimate failure loads. Variances of the ultimate failure loads were also compared using a Fischer test; $p < .05$ was considered significant. Due to the large scatter of the stiffness of fixation data, a statistical analysis of the data would require 100 samples. Therefore, comparison of the stiffness data was made only on qualitative grounds.

A.1.3 RESULTS

A.1.3.1 Ultimate Failure Load and Stiffness

A summary of the pullout data is shown in Table A.1. Typical load-elongation curves for both groups are plotted in Fig. A.4. Mean initial pullout force and 95% confidence limits for groups 1 and 2 were found to be 516 ± 37 N and 588 ± 37 N, respectively. The mean pullout force of group 2 was significantly higher than group 1 ($P = 0.003$). The standard deviations for groups 1 and 2 were 51 N and 52 N, respectively. The variances in the two groups were not statistically different. Mean stiffness and 95% confidence limits for groups 1 and 2 were 56.5 ± 10.2 N/mm and 52.1 ± 12.8 N/mm, respectively.

Table A.1: Maximum pullout force, pullout force range, stiffness, stiffness range, 95% confidence intervals, and standard deviations for the two fixation methods of soft tissue grafts loaded in-line with a porcine bone tunnel.

Fixation Method	RCI Screw	RCI Screw and EndoButton
	Group 1 (n=10)	Group 2 (n=10)
Maximum pullout force (N)	516	588
SD	51	52
Pullout force range (N)	472 – 629	525 – 702
95% confidence interval (N)	479 – 552	551 – 626
Stiffness (N/mm)	56.5	52.1
SD	14.3	17.8
Stiffness range (N/mm)	29.5 – 78.2	24.4 – 82.5
95% confidence interval (N/mm)	46.3 – 66.7	39.3 – 64.8

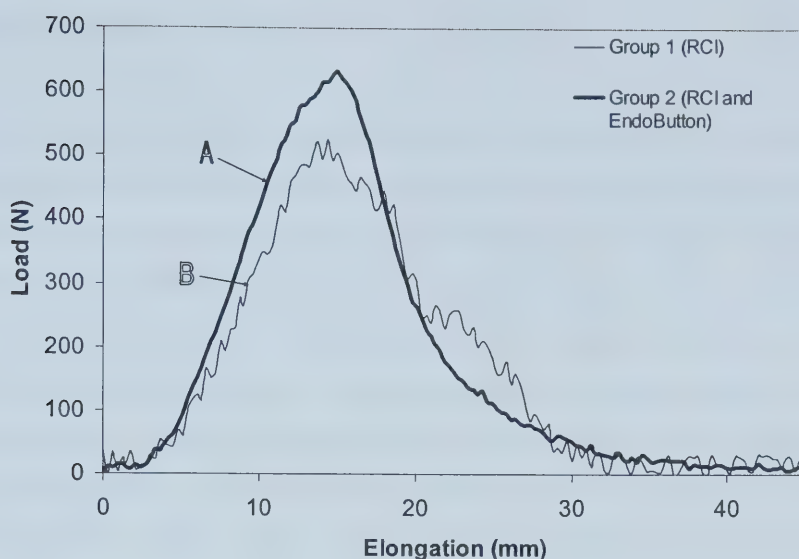


FIGURE A.4: Load-elongation curves of a porcine patellar tendon fixated with RCI screw (group 1), and RCI screw and EndoButton (group 2). (A) represents the slope (stiffness) of group 2; (B) represents the slope (stiffness) of group 1.

A.1.3.2 Failure Mode

Table A.2 summarizes the results of the failure modes. In group 1 failure at the attachment site was predominant, with 90% of the specimens failing from the graft sliding past the RCI screw. The interference screw remained in the femoral osseous tunnel after failure. One specimen failed by midsubstance graft failure. In group 2 all specimens failed in the same manner. The mode of failure was a combination of attachment site failure and tissue failure. As the soft tissue graft slid past the interference screw, the sutures tore through the graft. The interference screw remained in the bone tunnel and the knots on the EndoButton remained secure after failure.

Table A.2: Failure modes for the two fixation constructs.

Method	n	Graft sliding past RCI	Midsubstance graft failure	Suture tearing graft
Group 1 (RCI Screw)	10	9	1	N/A
Group 2 (RCI and EndoButton)	10	10*	0	10*

*For group 2 (RCI and EndoButton) mode of failure was a combination of both the graft sliding past the RCI screw and the suture tearing the graft.

A.1.4 DISCUSSION

The objective of the present study was to investigate the influence of the additional application of an EndoButton on the fixation properties of interference screws used for soft tissue grafts (i.e. a novel hybrid fixation for soft tissue). We hypothesized that combining EndoButton fixation with interference screw fixation would increase fixation strength and stiffness. Stronger and stiffer fixation techniques than those currently available are required due to increasingly aggressive rehabilitation techniques.

Although the testing materials did not reflect those of human anterior cruciate ligament reconstructions, it was nevertheless appropriate for the purposes of this study. A porcine patellar tendon, used to model a soft tissue graft, was inserted into porcine femoral bone. A previous study has shown that the material properties of the porcine patellar tendon are similar to the human patellar tendon¹³. A porcine bone model was used in this study because the biomechanical properties of porcine specimens are relatively uniform, whereas the biomechanical properties of human specimens (especially the more readily available elderly cadavera) are highly varied¹⁴. Bone mineral density has been found to be an important variable in terms of tendon interference fixation⁵. Miyata et al.¹⁴ have evaluated the fixation strength of metal interference screws in porcine bone. They have found that no substantial difference seems to exist between porcine bone and human bone with respect to ultimate strength of interference screw fixation. As the intention of this study is to compare fixation techniques, the use of the porcine model would be advantageous.

Two fixation constructs were compared for initial fixation strength, stiffness, and failure modes. The two constructs consisted of the soft tissue graft secured by RCI screw alone and the RCI screw combined with the EndoButton (using two Number 5 sutures). The failure load for grafts secured directly with the interference screw and periosteally with the EndoButton (group 2) was 588N. This initial fixation strength is significantly greater when compared to the initial fixation strength of grafts secured by interference screw alone (group 1, 516N). This 14% increase in pullout strength can be attributed to having two fixation points instead of one direct fixation point. When the graft fails, it is due to failure at both sites of fixation. Direct fixation with interference screw alone mainly fails

by the soft tissue graft slipping around the interference screw. By having two fixation points, a larger load is needed to fail the combined fixation construct than that needed to fail just the direct fixation construct. In terms of stiffness, the data show very similar values for group 1 (56.5 ± 10.2 N/mm) and group 2 (52.1 ± 12.8 N/mm).

Our data for the initial fixation strength of a soft tissue graft secured only by interference screw yield slightly higher loads and stiffness when compared to other studies using animal models (Table A.3). Comparisons were made only to animal studies because the bone mineral density of animal models has less variability than cadaver bone, thus the failure loads are more consistent. These studies¹⁰⁻¹¹ used hamstring tendons loaded in line with bovine calf tibial bone tunnels. The mode of failure was similar for all four studies listed in the Table. In our study, 9 out of 10 specimens failed by the graft pulling out of the bone tunnel. In comparison, the majority of the specimens tested by Weiler and associates¹⁰⁻¹¹ also failed in this manner. The construct combining three strand semitendinosus tendons secured by biodegradable interference screws produced a mean initial failure load (507 ± 93 N) and stiffness (57.9 ± 13.8 N/mm)¹⁰. These results are very similar to those of this study when interference screws alone were used. The other two constructs produced lower failure loads.

Table A.3: Comparison of failure load and stiffness of pullout studies using animal bone. Values shown for failure load and stiffness are Mean $\pm$ SD.

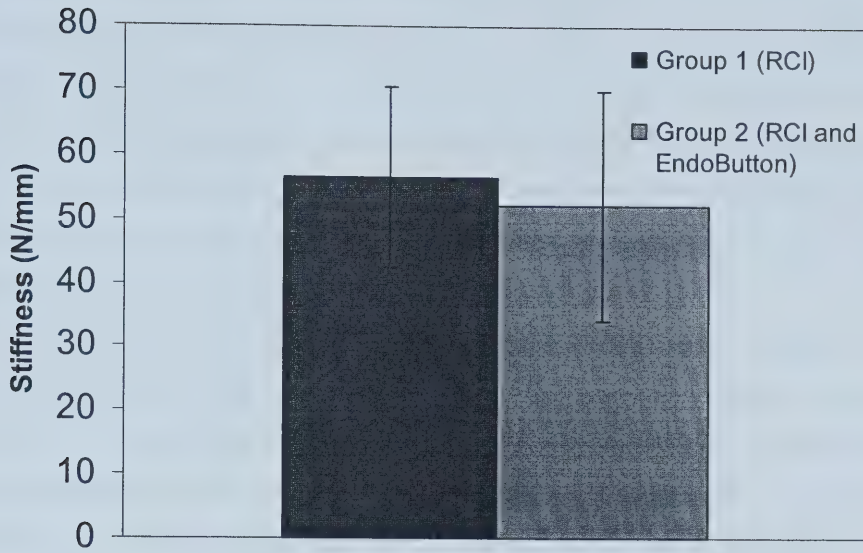
Bone type	Fixation device	Rate of applied force	Grafts	Failure load (N)	Stiffness (N/mm)
Calf tibial bone ¹¹	Biodegradable screw	1 mm/sec	Quadrupled semitendinosus and gracilis tendon	385.9 $\pm$ 185.6	25.69 $\pm$ 8.47
Calf tibial bone ¹⁰	Biodegradable screw	1 mm/sec	Three strand semitendinosus tendon	507 $\pm$ 93	57.9 $\pm$ 13.8
Calf tibial bone ¹⁰	Titanium Interference screw	1 mm/sec	Three strand semitendinosus tendon	419 $\pm$ 77	39.7 $\pm$ 10.9
Present study, porcine femoral bone	Titanium interference screw	3 mm/sec	Tubularized porcine patellar tendon	516 $\pm$ 51	56.5 $\pm$ 14.3

One reason for ostensibly higher loads observed in this study may be due to the relatively faster loading rate used to test our specimens. Table A.4 displays three studies which all used metal interference screws to secure grafts to older cadaveric femoral bone tunnels. Loading was done in line with the tunnels. The table lists the loading rate used and the associated failure load of the constructs. The data in Table A.4 suggest that the increased loading rates could be a factor in the increased failure loads observed. Extending this to the animal studies of Table A.3, this may explain the observed higher failure load in comparison to those studies¹⁰⁻¹¹. However, one also has to recognize the differences between the systems studied.

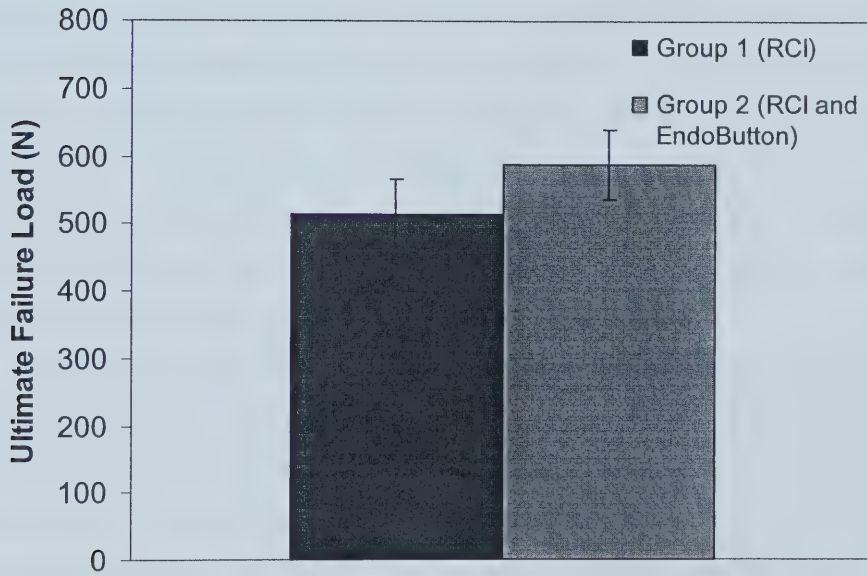
Table A.4: Comparison of ultimate loads of pullout studies using elderly cadaver bone under various loading rates. Values shown for ultimate load are Mean $\pm$ SD, except for the bottom value where only the Mean is shown.

Mean cadaver age	Graft	Rate of applied force	Ultimate load
69 years ⁸	Quadrupled semitendinosus and gracilis tendon	0.33 mm/sec	242.0 $\pm$ 90.7 N
79 years ¹⁵	Patellar tendon	1 mm/sec	322 $\pm$ 178 N
62 years ⁹	Patellar tendon	4.2 mm/sec	436 N (range: 111 – 903 N)

Fixation stiffness allows loading on the graft to increase without an increase in the slackness of the graft. Thus, a stiff fixation will reduce the laxity of the knee¹⁶. It appears from Fig. A.5a that the mean stiffness values of groups 1 and 2 are very similar (especially when considering the error bars). A separate unpublished study done by our group found the stiffness of fixation by EndoButton alone to be 52% less than fixation by RCI screw and EndoButton. Thus, in combining the two modes of fixation, the stiffness of fixation is provided mainly by the interference screw and not by the EndoButton.



(a)



(b)

FIGURE A.5: (a) The mean stiffness of fixation and (b) the mean ultimate failure load of the two fixation constructs. The error bars represent the standard deviations.

Similar standard deviations for the ultimate failure loads in groups 1 (51 N) and 2 (55 N) indicate a uniform distribution of fixation strength among the two groups (Fig. A.5b). The standard deviations are also very similar for the stiffness of fixation of the two groups. Clinically, this means that using femoral-ended periosteal fixation in conjunction with direct fixation does not reduce the reliability of direct fixation itself in terms of ultimate failure load. From a mechanical point of view, no additional danger of failure is anticipated when combining both methods of fixation.

As this is an animal study, we must recognize that the absolute values shown in this study are not strictly equivalent to those obtained from human femur-graft complexes. As a result, the qualitative characteristics of combining the interference screw with the EndoButton rather than the numerical values are of interest. Nevertheless, the results hold the promise of potential advantages gained by using the suggested novel hybrid fixation method.

Also, it should be noted that the complexes were loaded in line with the tunnel. This worst-case scenario was used to examine the specimens under extreme loading and was not intended to focus on daily physiologic motions. Forces on the fixation sites during physiologic motions are expected to produce lower pullout loads than in worst-case scenarios⁵. In weight bearing positions, the loading is different than in this worst-case scenario, as loading at the femoral side occurs at an angle. However, it is anticipated that combining the interference screw with the EndoButton will still have a positive effect on fixation in physiologic motions.

There are clinical situations where a closed loop device cannot be used (i.e. the free end of quadriceps tendon grafts and Achilles tendon allografts). Revision surgery is sometimes complicated by poor bone density and tunnel-graft mismatch. In these situations, when fixing the soft-tissue end of the graft, a construct of interference fit screw with EndoButton and sutures could be used to improve graft fixation. The data presented in this study show a definite motivation for using both direct and periosteal fixation. The use of both forms of fixation provides a higher ultimate failure load than a construct using interference screw fixation alone, without affecting the stiffness in any noticeable way. In order to gain the most benefit from the two methods of fixation, the interference fit screw should be inserted next to the tendon in such a way that the failure

mode occurs through slippage and not by tearing of the graft. If tearing occurs, the effect of the periosteal fixation is lost.

For clinical situation where a closed loop device can be used, a new closed loop EndoButton has been developed for use with hamstring tendons (Smith and Nephew Endoscopy, Mansfield, MA). In the future, we would like to examine the effects of this closed loop EndoButton when combined with an interference screw

A.1.5 CONCLUSIONS

It was found that providing periosteal fixation using an EndoButton with two Number 5 sutures in conjunction with direct fixation using an interference screw provides an increase in initial fixation strength over direct fixation by itself. The stiffness of the combined construct is similar to that of a soft tissue graft fixated by interference screw alone. It is anticipated that this novel hybrid fixation method will provide increased fixation strength for soft tissue grafts, allowing the accommodation of the current rehabilitation procedures. A strategy of combining direct and periosteal fixation may hold the key to improved fixation characteristics.

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